K.-U. Schmitt
P. Niederer
M. Muser
F. Walz

Trauma Biomechanics

Accidental injury in traffic and sports

2nd Edition

Springer

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With 88 Figures and 30 Tables

Preface

Everyday, more than 140’000 people are injured, 3000 killed, and 15’000 disabled for life everyday on the world's roads. Likewise, sports related injuries are numerous and have a significant socio-economic impact. The field of trauma biomechanics, or injury biomechanics, uses the principles of mechanics to study the response and tolerance level of biological tissues under extreme loading conditions. Through an understanding of mechanical factors that influence the function and structure of human tissues, countermeasures can be developed to alleviate or even eliminate such injuries.

This book, Trauma-Biomechanics, surveys a wide variety of topics in injury biomechanics including anatomy, injury classification, injury mechanism, and injury criteria. It is the first collection I am aware of that lists regional injury reference values, or injury criterion, either currently in use or proposed by both U.S. and European communities. Although the book is meant to be an introduction for medical doctors and engineers who are beginners in the field of injury biomechanics, sufficient references are provided for those who wish to conduct further research, and even established researchers will find it useful as a reference for finding the biomechanical background of each proposed injury mechanism and injury criterion. As more people become aware of and understand this subject, it will someday lead to better mitigation and prevention of automotive adn sports related injuries. I like this book very much and believe that you will find the same.

King H. Yang
Professor of Biomedical Engineering
and Mechanical Engineering
Director of Bioengineering Center
Wayne State University

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Kai-Uwe Schmitt, Peter Niederer, Markus Muser, Felix Walz

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1 Introduction

The human being is exposed to mechanical loads throughout his or her life. Besides gravity and forces due to electromagnetic fields, there is a great variety of forces acting on the human body from contacts with the surrounding; in addition, numerous forces are generated in the course of physiological processes inside the body in the different organs and tissues. Throughout evolution, all forms of life adapted their physiology to mechanical interactions; some of them to the extent that a proper function in fact requires the influence of forces, for example bone remodelling.

The science of biomechanics is devoted to the analysis, measurement and modelling of the effects which are observed under the various mechanical loading situations primarily in humans, but also in animals and plants. As this definition suggests, a quantitative approach is thereby in the foreground. The range of forces which is of interest is enormous: Internal forces may originate from the action of molecules, contractile fibres on a cellular level or muscles on a macroscopic scale, moreover, pressures and shear stresses may be generated by biological fluid flows or active biological transport processes including osmosis. External forces, in turn, occurring in everyday life may span a virtually unlimited extent. Accordingly, the forces of interest in biomechanics cover typically a range from pN to MN (lower or higher forces, respectively, are hardly considered because of lack of biological effect on the lower side or complete devastation on the upper), and they may vary with time within picoseconds to years.

An inevitable consequence of forces acting in- or outside the human body consists of the possibility that they may cause injury. Such adverse consequences are usually associated with the action of excessive external forces impinging during unfavourable events with which we may be confronted in daily life. Internal forces, in contrast, are mostly thought to be governed by anatomical or physiological constraints which prevent the occurrence of injury. Yet, broken ribs due to intense coughing, rupture of muscle fibres because of tetanic contraction or endocardial bleeding in

cases of hypovolaemic shock are injuries resulting from forces produced by the body itself.

The special discipline of biomechanics which is concerned with injury caused by mechanical interaction is denoted as biomechanics of injuries or trauma-biomechanics and is the subject of this book. Since there are a great many types of injuries, injury mechanisms and activities which are prone to cause injury, a large variety of human activities and situations where excessive loads may occur have to be considered. It becomes thereby evident, that trauma-biomechanics is strongly dominated by its interdisciplinary character. First of all, the field of biomechanics itself covers a wide range of areas of interest, from macroscopic motion analysis in sports, for example, to the sub-microscopic modelling of molecular transmembrane transport. The knowledge obtained in such different fields greatly contributes to trauma-biomechanics' work, in that for an in-depth understanding of injury processes all aspects from the macroscopic scale to the sub-cellular level may have to be taken into account. Therefore, many subjects of importance for trauma-biomechanics relating to basic mechanics, anatomy and physiology have to be covered in order to be able to treat the entire field to some comprehensiveness.

1.1 About the contents of this book

Several preliminary remarks are useful in order to elucidate the extent and limitations of the subjects treated in this book:

1. A distinction has to be made between injury resulting from unexpected, sudden and single events, i.e., accidents in a strict sense, and injury caused by the chronic exposure to unfavourable loads over extended periods of time. A head injury of a pedestrian which is sustained from an impact on the hood of an automobile during a collision, or the gradual destruction of hair cells in the inner ear as the result of a chronic exposure to loud music - both examples are associated with injury, yet, the type of injury, injury mechanisms, tolerance levels, injury criteria, reconstruction and analysis methods as well as protection measures differ basically. Also with respect to insurance and liability issues, procedures are greatly different.
2. The injury causing period in the course of a traffic accident has duration of 100 msec - 200 msec typically whereby the early part of this period is often decisive. In many cases the person involved is not aware of the event and does not (cannot) react prematurely to the imminent

danger. Accordingly, muscular reactions which set in with a time delay of 60 msec - 80 msec are often of secondary importance only and can be disregarded. The situation is basically different in case of chronic overloading, where physiological and also psychic reactions are always in the foreground.

3. A further important aspect is related to age. Humans change their mechanical properties, in particular injury tolerance, decisively during aging towards unfavourable levels. There are a number of reasons for this, among them, a reduction of tissue compliance due to a decrease of body water content along with an increase of collagenous cross linking, furthermore, a demineralization of bone. As a result, the advent of injury, primarily bone fractures, increases dramatically with age. The incidence even of spontaneous fractures, occurring under normal physiological loads, is well known. In view of the aging population in the industrialized countries, this aspect has to be given particular attention.
4. The mechanical response of the body in case of pathological alterations may furthermore be significant. Renal trauma as a result of stress concentration around a cyst has been observed in urology or the aggravation of the effects of a whiplash-type event due to preexisting neck impairment is a well known complication.
5. Under very restrictive conditions, however, microinjury on a cellular level may to some extent be advantageous for tissue regeneration. Figure 1.1 shows the microcallus formation following microdamage in spongy bone which may serve as an example of an injury which stimulates bone remodelling. After a long and strenuous hiking tour such microinjuries in the bones of a healthy foot are quite common. Chronic overexposure, in contrast, may lead to quite adverse developments. Figure 1.2 shows a marathon runner whose skeleton was largely demineralized due to excessive training.

Fig. 1.1 Microcallus formation. The picture exhibits a 3D micro computed tomography (micro CT) scan of the excised portion (biopsy) of a human iliac crest where microfractures induced new bone formation [courtesy: Prof. R. Müller, ETH Zurich].

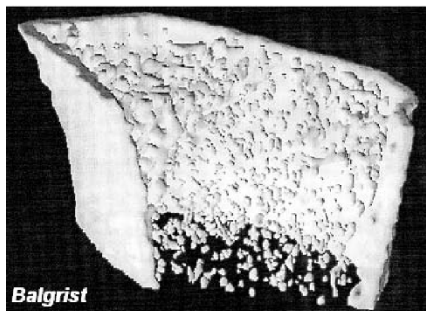



Fig. 1.2 28 year old woman (left) and Micro-CT scan (non-invasive) of radius close to the wrist (right). The extreme demineralisation of the radius is to be attributed to excessive training as a Marathon runner [Courtesy: Prof. M. Dambacher, MD, Balgrist Orthopaedic Hospital, University of Zurich].

6. Injuries are suffered mostly in connection with motion (sports, household, etc.) or mobility (traffic). While in general biology the use of animal models (under restrictive regulations) is widespread, the nonlinearity inherent in motions and related injury mechanisms prevents scaling up from, say, rats to humans almost completely. Accordingly, except for aspects of basic physiology, only scarce information in trauma-biomechanics derives from animal experiments.
7. When the entire spectrum of "injury" including causation, frequency, prevention, mitigation, rehabilitation, long-term sequelae and socio-economic consequences is considered, clinical medicine cannot be disregarded, since the treatment of injury is made by medical doctors providing in- and outpatient service. It is thereby often forgotten that the overall reduction of specific mortality (i.e., mortality by case) which is observed in most activities associated with a risk of injury, is partly due to dramatic developments of emergency rescue services, first aid procedures and intensive care treatment. A drawback however is that an analysis of injury mechanisms and accidental events is sometimes made by physicians without a complete knowledge of the relevant facts. This may be a consequence of their intense and highly demanding work with patients. Yet, an objective assessment of the severity and causality aspects of accidents resulting in injuries requires a multidisciplinary approach. In addition to the medical information retrieved by clinicians, all technical and biomechanical circumstances have to be taken into account in accident analysis. This is especially

important in cases of forensic expert witnessing. A specialized education and extensive experience is required for this purpose.

8. This book is limited to the mechanics of injuries which are inflicted without intention. However, injuries may also be conveyed intentionally in a criminal, terroristic, or battlefield environment. Related subjects include primarily wound ballistics, protective garment for soldiers or low-injury producing police weapons. The reader who is interested in such issues is referred among other to several publications by the Int. Committee of the Red Cross (http://www.icrc.org) where also further references can be found. The overall significance of intentional injuries should in any event not be underestimated; e.g., there were around 12'000 deaths in the US (2001) due to the use of firearms (not including accidental firearm incidents) in comparison with some 11'000 persons who were killed in the same year as passengers of an automobile (drivers not included). Nontechnical aspects (social, political, psychological, general society-related) are particularly important in this context.
9. The least serious injuries are of course those which do not occur. Accordingly, injury prevention is given a high priority in all situations where injury may happen. The prevention of traffic accidents has been recognized and implemented as an important governmental task since decades. In contrast, injury prevention in sports has primarily been perceived by international and national sports associations within the framework of sports medicine, mostly in the form of rigorous reglementation, ban of certain particularly violent variants of sports, development of protective devices such as helmets or shin guards and trainer education. Insurance companies furthermore support all injury prevention campaigns as part of their mission whereby these efforts include especially also workplace and household accidents. While all of these preventive activities are oriented towards pre-accident conditions, after injury has occurred and healed, extensive rehabilitation is often required. Again, government agencies, sports federations, professional work associations, clinical medicine as well as insurance companies develop extensive efforts. Since this book is devoted and limited to trauma-biomechanics, aspects of prevention and rehabilitation are only included in so far as there is a direct relation with the incurrence of injury.

Most systematic and quantitative research in trauma-biomechanics has been made in connection with traffic accidents, although injuries sustained in sports, at the workplace or during household activities are likewise

prominent. There are mainly two reasons for this. First, more serious and fatal injuries are sustained and higher social costs are involved in traffic accidents than in other human activities as will be discussed in this chapter. Accordingly, liability problems along with political interventions and government rulemaking put the automobile industry under enormous pressure (the public response to the 1965 book by Ralph Nader, "Unsafe at Any Speed" was overwhelming) and stimulated comprehensive research and development activities. Second, although traffic accidents, like all other types of accident, exhibit a wide variety and variability, it is nevertheless possible to identify some "typical" or "representative" types of accidents, e.g., a frontal collision of an automobile against a barrier or a 90° side impact which lend themselves for a well defined testing protocol and quantitative analysis. In a sports, workplace or household related environment it is in contrast difficult to find typical situations which cover a substantial amount of injury-producing situations.

In comparison with traffic accidents, the literature on injury in sports, etc. is - although abundant - less stringent from a biomechanics' point of view in that general statistics, mostly descriptive explanations of injury mechanisms, medical treatment strategies and practical recommendations for trainers or people responsible for workplace safety dominate over quantitative analysis. Whenever quantitative information is sought with respect to human tolerance or injury criteria derived thereof, the literature on sports injuries refers almost exclusively to results from traffic accident analysis. It is also remarkable that most investigations on sports accidents are made in those disciplines which are associated with an enormous financial background such as soccer, American football or skiing, while less prominent areas, e.g., orienteering, receive much less attention.

An even greater variety from an anatomical and physiological point of view in comparison with accidents exists with respect to the occurrence of injuries due to chronic mechanical (over-) exposure. A distinction between impairment due to chronic exposure and diseases which are unrelated to the exposure is often difficult or impossible. Psychic influences are particularly important. Quantitative information is scarce. Vibrations of construction machinery, for instance, or noise levels in factories and entertainment facilities are limited by regulations which are derived from long-term statistical evidence rather than from physiological experiments.

From the above considerations it becomes understandable that the backbone of this book is essentially based on the trauma-biomechanics of traffic accidents. After a general chapter describing basic definitions and methods, a sequence of chapters is presented that deal with the different body regions. These chapters are arranged systematically by starting with a

brief outline of the anatomy of the body region in question, limited to those aspects which are of special importance in view of injury mechanisms. Furthermore, the range of possible injuries, underlying injury mechanisms and the biomechanical response to loading of this body region are described. Known injury tolerance values and injury criteria based thereon to assess the likeliness of injury are discussed. Following injuries sustained in traffic accidents, sports injuries are treated whereby selected special aspects of anatomy relevant for the understanding of related injury mechanisms, injury analysis and tolerance criteria are included. An important part in each chapter is furthermore devoted to protection measures which are recommended or regulated in order to mitigate injuries. Where appropriate, additional information on injury prevention measures or other special subjects is presented. For further reading, references are given at the end of each chapter.

A short section on injuries due to chronic exposure to mechanical loading is added at the end. Many practical issues in connection with such injuries fall within the framework of ergonomy, general workplace safety and management of occupational safety hazards, however. For example a thromboembolism occurring during a long air voyage is a problem of ergonomic seat design and passenger behaviour rather than of trauma-biomechanics. As this book is limited to the latter field, such subjects are not included. E.g., over the home page of the US Occupational Safety and Health Administration (http://www.osha.gov) relevant information can be found.

1.2 Historical remarks

Biomechanics as a science is as old as mechanics. While e.g. Giovanni Alfonso Borelli (1608 - 1679) devoted much of his time to the analysis of bird flight and swimming of fishes, Leonhard Euler (1707 - 1783), the creator of continuum mechanics, wrote an extensive treatise on the principles of the motion of blood in the arteries ("Principia pro motu sanguinis per arterias determinando", op.posth.). Until the mid 19th century, however, the mechanics of injury or trauma-biomechanics was not the subject of systematic research. This might be attributed to the fact that dangers were ubiquitously imminent and injury must have been considered a natural feature associated with life. The reader should not forget that through 2000 years of history up to 1945, there has never been a period longer than 15 years without war in Europe! Injury prevention was rather

straightforward and pragmatic, e.g. in the form of cuirasses for knights.

The first known systematic and scientific approach towards trauma biomechanics was taken by the German anatomist Otto Messerer in Munich who published his results in the year 1880 (Cotta Publ., Stuttgart) under the heading "On the Elasticity and Strength of Human Bones" ("Über Elasticität und Festigkeit menschlicher Knochen"). His activity was however rather isolated at that time. Nevertheless, the "Messerer-wedge" is well known in forensic science and still serves as a reminder for his seminal work.

As mentioned before, the field of trauma-biomechanics today is mostly centred around injuries sustained in traffic accidents. Yet, historically, their roots are in aviation. During the 1st National Conference on Street and Highway Safety (USA 1924) simple and practical aspects of traffic safety such as the colours of traffic lights or driver education dominated while biomechanics was not (yet) of concern. In contrast, trauma-biomechanics were already at that time a significant issue in the field of military aviation where the human body is exposed to extreme mechanical loading conditions. After having observed many accidents with aeroplanes, Hugh DeHaven, who can be considered as the "Father of Trauma-Biomechanics", started an analysis of the underlying injury mechanisms. In 1942, he published a first work titled "Mechanical Analysis of Survival in Falls from Heights of 50 - 100 Feet". In the following years, military aviation remained the focus of research into trauma-biomechanics. Transsonic flight and ejection seat dynamics were among the problems that stimulated this research. Basic experimental methods like subtraumatic volunteer experiments to determine the biomechanical response of the human body or the development of anthropomorphic test devices (dummies) followed.

Probably the most famous pioneer in aviatic trauma-biomechanics was Colonel John Paul Stapp. He became particularly well-known for his experimental work, including several experiments subjecting himself to various impacts. In one of the most spectacular series of these experiments conducted during the early fifties, Stapp was seated on a rocket sled. From a velocity of approximately 1000 km/h, the sled was stopped within 1.4 s in a water bath resulting in a sled deceleration of 40 times gravity (Figure 1.3). No serious injuries were reported from this experiment. Stapp, whom the Time Magazine called "the fastest man on earth and No. 1 hero of the Air Force" (Time, September 12/ 1955), also founded an annual conference which still is one of the leading platforms for the discussion of trauma-biomechanics' related subjects - the Stapp Car Crash Conference. Stapp died in 1999 at the age of 89 years.

Later on, astronautics necessitated the investigation of human

Fig. 1.3 Colonel Stapp sitting on the rocket sled "Sonic Wind No. 1" with which he was subjected to a deceleration of approximately 40g [http://www.stapp.org].

physiology under totally opposite conditions as considered here, namely weightlessness. There were nevertheless developments which were also of interest in trauma research. E.g., the first computer model for the simulation of 3D human motion (R.D. Young, Texas A&M, 1970) was developed in connection with the analysis of human motion patterns in case of absence of external forces. With respect to traffic accidents, McHenry (Calspan Corp., Buffalo) wrote the first computer model for motions of humans involved in a frontal crash. Since in this case the influence of external forces is of importance, a large portion of the simulation was devoted to the modelling of interactions of body parts with surrounding structures. As this complicated the computational and numerical complexity decisively, first models were limited to planar (2D) motions.

During the early days of automotive transportation, safety issues were primarily considered to be a domain of the driver who was assumed to be responsible for driving in a manner that would safeguard drivers, passengers as well as occupants in other vehicles, bicyclists and pedestrians. Restraint systems were thought of (Figure 1.4) but not widely

implemented until after World War II. Nevertheless, during the 1920s and 1930s, car manufacturers gradually improved the vehicle design also with respect to safety. Reliable and durable four-wheel breaking systems were introduced and laminated safety glass replaced the plate glass that was used for windscreens. Further developments focused on lighting such as sealed-beam head lamps and the wheels by introducing tubeless tires. All-steel car bodies were used instead of wooden structures and thus increased the stiffness of the vehicles.

It was the consequence of the rapidly increasing mobility after World War II along with a dramatic increase in numbers of injuries sustained in traffic accidents that in-depth emphasis was finally given to these problems. The Automotive Crash Injury Research programme (ACIR, Cornell University, 1951) represented the first systematic approach with respect to injury analysis in traffic accidents. An important development set subsequently in when the concept of a stiff passenger compartment combined with a defined crush zone was realised. The steering column as a possible source of injury received at the same time attention, too, leading to the development of multi-element and energy-absorbing steering columns. Further improvements included the crashworthiness of the instrument panel, the development of restraint systems like the three-point belt and the airbag. Furthermore, the terms "passive" and "active" safety were

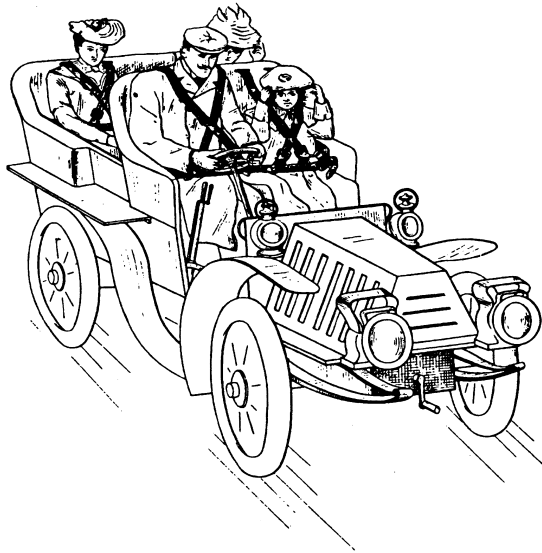



Fig. 1.4 The seat belt patent (1903) by Gustave D. Lebau.

established and systematic crash testing along with numerical simulation was started by automobile manufacturers. A comprehensive overview over the research made in automotive safety up to 1970 can be found in the 1970 International Automobile Safety Conference Compendium (published by SAE, New York).

Crash injury management, i.e., passive safety can be approached from three different levels. First, a reduction of injuries can be achieved by improved crashworthiness of the vehicle. This includes in the first place the design of energy absorbing structures. Second, the occupant motion in case of impact can be controlled. Restraint systems like the seat belt emphasise this aspect by keeping the occupant in the designated area and also connecting the occupant motion with that of the vehicle. Third, the actual impact, i.e. the contact between the human body and its environment, is addressed by way of energy absorption and distribution of the impact load over a greater contact area. Active safety, in turn, is a matter of braking systems, vehicle handling properties, pre-crash control (distance radar), and in an increasing fashion, computer-controlled driver assistance.

In addition to such technical advancements, also governmental bodies became increasingly aware of the safety issue in road traffic after World War II. By introducing first programmes for driver education, enacting driving regulations and developing new highway concepts, a higher safety standard was aimed at. Road planning and construction being furthermore within the responsibility of government agencies, management of accident risk by a favourable lay-out of the road environment, general traffic and speed control, guard rails, etc. is an important part of their task.

While to a great extent trauma-biomechanics' activities during the last 30 - 50 years have been concerned with injuries sustained in automotive accidents, also pedestrian accidents as well as accidents involving 2-wheel vehicles (Figure 1.5) are of increasing concern. However, as mentioned before, the traffic environment is only one field in which accidents occur. Injuries sustained in accidents at work, at sports or during daily life activities are also of importance. In an industrialised country (USA 2002), the number of non-traffic related accidental deaths was in fact higher than fatalities due to motor vehicle accidents (Table 1.1).

In Table 1.2 the number for accidents causing injury in sports (without specification of severity) are given for the USA (2007) together with the corresponding numbers for Switzerland; a comparison which highlights on the one hand the situation in a large and small industrialised country, but which also reflects the enormous differences between countries due to local habits and preferences on the other. The large number of injuries occurring

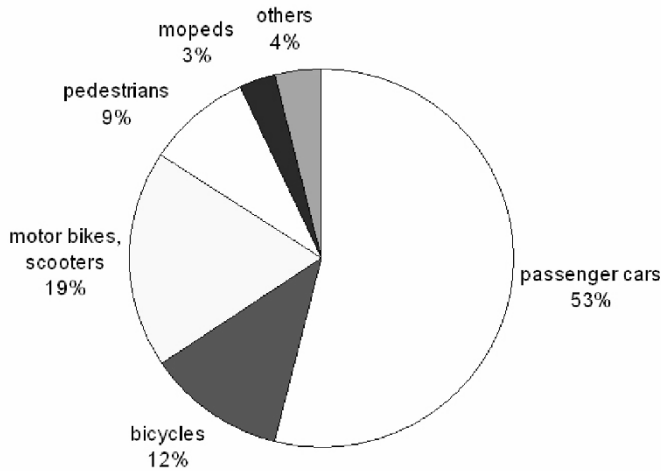



Fig. 1.5 The distribution of casualties recorded by the police in traffic accidents in Switzerland in the year 2005 [Swiss Council for Accident Prevention, 2006]. Occupants of passenger cars are injured most often.

Table 1.1 Causes of reported accidental deaths in the USA 2002 (Source: Nat. Vital Statistics Report, Vol. 50, Nr. 15, 2002). The total number of reported fatal incidents (persons) was about 98'000.

Cause of Fatal Accident	Percentage
Motor vehicle	44.3%
Slips, trips, falls	17.8%
Accidental poisoning (solid/liquid)	13.0%
Drowning	3.9%
Fires, burns, smoke	3.4%
Medical/surgical complication	3.1%
Other land transport	1.5%
Firearms (accidental)	0.8%
Other (nontransport)	12.2%

Table 1.2 Number of reported sports-related injuries in the USA (Source: Charles W. Nuttall, 5th Int. High Energy Physics Laboratories Technical Safety Forum, SLAC, 2005) and Switzerland (Source: Swiss Council for Accident Prevention, 2005). Note that favorite sports differ in the two countries: While American football and baseball are hardly played in Switzerland, soccer and skiing are quite popular.

Sport	USA, 1997	Switzerland, 2003
Basketball	644’921	5880
American football	344’420	na
Baseball, softball	326’569	na
Soccer	148’912	55’040
Trampolines	82’722	na
Skateboards	48’186	10’330
Golf, golf carts	47’777	na
Skiing	na	49’660
Snowboard	na	28’890
Sled, bobsleigh	na	10’8000

in sports furthermore indicates that the vast majority of these injuries are not life threatening.

Considering worldwide statistics, it is nevertheless found that traffic accidents account for the highest number of fatalities: While the World Health Organisation (WHO) estimated a total of 1.2 Mio. deaths from traffic accident worldwide for the year 2002, the International Labour Organisation (ILO) extrapolated the number of fatal work accidents as 335'000 in 1998. The number of fatalities per time spent for certain activities may serve as an indicator for the danger associated with these activities. Table 1.3 demonstrates that traffic participation per se is not particularly dangerous in comparison with other activities (to a great extent due to the enormous efforts made in traffic safety), yet, this effect is by far surpassed by the time spent in traffic.

Table 1.3 Estimated FAR (Fatal Accident Rate, defined as number of deaths per 108 hours exposed activity) for various activities (Source: Charles W. Nuttall, 5th Int. High Energy Physics Laboratories Technical Safety Forum, 2005).

Acitvity	FAR
Travelling by train	5
Travelling by car	57
Skiing	71
Cycling	96
Motorcycling	660
Canoing	1000
Professional boxing	7000

2 Methods in Trauma-Biomechanics

Work in trauma-biomechanics is subjected to a number of limitations which are less stringent or even totally absent in other fields of the technical and life sciences. First of all, experiments involving loading situations with humans which are prone to cause injury are excluded. Second, animal models are of limited use because of the difficulty to scale trauma events reliably from animals up or down to humans. A number of experiments in connection with seat belts were nevertheless made in earlier years with pigs [Verriest et al. 1981] since their thorax resembles the human thorax mechanically to some extent; likewise, monkeys were subjected to impact in order to study head motion and neck dynamics [Ewing et al 1978]. Anaesthetised animals provide moreover a model to investigate physiological reactions at high mechanical exposure levels. Questionable representativeness with respect to human biomechanics in spite of some similarity, furthermore, cost and above all ethical considerations along with public awareness limit however such experiments to special circumstances today.

Accordingly, methods applied in trauma-biomechanics are to a great extent indirect and include mainly approaches based on

- statistics, field studies, data bases (2.1)
- injury criteria, injury scales and injury risk (2.2)
- basic mechanical concepts and accident reconstruction (2.3)
- experimental models (2.4)
- impact tests performed in the laboratory (2.5)
- numerical simulation (2.6)

2.1 Statistics, field studies, databases

Epidemiology is of fundamental importance in trauma-biomechanics and it represents also the oldest methodological approach. The identification of

injury risks and the analysis of causative factors are largely based on epidemiologic evidence which in turn stimulates the development of intervention strategies as well as of technical and legal countermeasures with the aim of accident prevention and injury reduction. Whether such countermeasures are indeed effective can again only be decided on the basis of statistical surveys which often require long-term studies. Hence, when working in the field of trauma-biomechanics, in particular towards issues related to injury mitigation and prevention, the acquisition and in-depth analysis of real world accident data is an indispensable prerequisite and research tool.

The collection, classification and interpretation of accident data have to be subjected to a careful assessment with respect to the sampling process in that in most cases the available data set is not exhaustive but is limited to a selected sample. One should always be aware of the fact that major limitations on the applicability of the results of any statistical evaluation are already incorporated in decisions on how and what data are collected. In contrast to fully controlled laboratory experiments, uncertainties arise for example due to the circumstance that many important parameters in real accident situations are not monitored and may exhibit a large variability. In addition, the memory of those involved in an accident or acting as witnesses may be inaccurate about the details or influenced by legal or insurance related considerations. Other factors such as the current composition of the vehicle fleet in case of traffic accidents, the price of gasoline, changes of legislation, adaptation of rules in contact sports, or changes with respect to insurance coverage of workplace accidents have to be considered when attempting to analyse the influence of newly introduced safety measures. A sound statistical evaluation may also fail because of an insufficient number of cases available for a representative analysis.

With respect to methodology, two types of accident data bases or injury surveillance systems can be distinguished, viz., general accident collections involving a large, possibly complete coverage of accidental events on the one hand, and in-depth studies of selected cases on the other. General large-scale accident files are typically collected by the police, other government bodies or insurance companies and are presented in annual accident statistics. They usually contain a large number of cases but only limited information per case. In turn, in-depth case analyses are performed by specialised teams which attempt to recover as much detail as possible of each case under scrutiny - which somewhat cynically can be regarded as an involuntary experiment - on the basis of investigation of the accident scene, workplace or household locations and installations, vehicles, sports

accessories, furthermore, police reports, witness depositions, interviews, medical records, weather reports, video coverage of sports events and on-site reconstruction with original vehicles or installations. Numerical simulation is then often applied to elucidate loading conditions and to relate them with injury patterns. Needless to say that such investigations are associated with a high expense and only a limited number of cases can be evaluated in this fashion. Representativeness is a particularly critical aspect in this approach.

Insurance companies often have larger collections than governmental bodies because accidents are reported to insurance companies for financial reasons while more reluctance is present with respect to involving the police, in particular in case of self-accidents without the involvement of a second party. Yet, insurance data are often not accessible, and if yes, not detailed or biased.

Cases included in large-scale data collections are moreover often not collected and analysed by specialists in accidentology and may contain significant errors and be selected according to criteria which are not applied uniformly. Accordingly, the results obtained from different data bases are often difficult to compare due to differences in the data collection schemes. Even within one specific data base type, e.g. police records, differences in basic definitions, data set volume or privacy policies may vary considerably from source to source. Whether e.g. an elderly patient who dies in a hospital from pneumonia two weeks after a severe traffic accident is indeed a traffic accident victim and included in the statistics or not may depend simply on the reporting practice of the hospital.

In most industrialized countries, accidents associated with traffic, workplace, household and sports fall within the competence of different government agencies, foundations, private institutions, sports associations, insurance companies, etc. with little mutual interaction. Reporting and investigation practises may differ along with injury prevention strategies such that comparisons between various types of injury-producing circumstances have to be made with great care. Uniform statistics are mostly available from small countries like Switzerland where the Swiss Council for Accident Prevention (bfu) provides a comprehensive coverage of accident data.

The largest systematic collections and statistics on traffic accidents are provided by the US National Highway Traffic Safety Administration (NHTSA). They include general data with respect to vehicles, crashworthiness and trends (National Automotive Sampling System, NASS) as well as information on traffic fatalities in the Fatal Accident Reporting System (FARS). An overview over these activities can e.g. be

found in Compton (2002). Similar, although sometimes less systematic information is available from most other countries worldwide.
Work place safety issues are comprehensively addressed in the statistics of the US Occupational Safety & Health Administration (OSHA). In most industrialized countries, furthermore, workplace accidents are covered by government controlled insurance organisations. General statistics are regularly available from such sources.

The situation with respect to sports accidents and injuries is somewhat different. Sports activities are largely voluntary and leisure-based (with the exception of mandatory participation in schools), are mostly covered by special insurance programs (in particular when competitive events or contact sports are involved), and product liability is highly diverse and selective (e.g., trampolines, diving boards in swimming pools, American football helmets, ski bindings). Specific, let alone general statistics involving comprehensive coverage over years, e.g. to analyse trends are largely missing. General awareness with respect to sports injuries has only recently increased. The Olympic Committee established in 1990 a Medical Commission and Library involving a Special Collection of Sports Medicine and Sports Science where the injury problem is partially included. While the Fédération Internationale de Football Association (FIFA) releases no systematic information with respect to soccer accidents and injuries, the Fédération Internationale de Ski (FIS) and the Oslo Sports Trauma Research Centre NSS announced in 2006 that they have agreed to develop an Injury Surveillance System (ISS) for the FIS disciplines of Alpine Skiing, Cross-Country Skiing, Ski Jumping, Nordic Combined, Freestyle Skiing and Snowboarding.

In-depth case studies are made by specialised teams, usually with a specific aim or involving a limited geographical area. In order to be useful, such efforts have to be maintained over years and a sufficiently large number of cases has to be collected observing uniform procedures. Most projects of this type which are documented in the literature are performed in connection with traffic accidents. For example, a team working at the Medical University Hannover (Germany) has been collecting data of collisions occurring in the area of the city of Hannover over many years. Because the data have been gathered systematically and following a uniform protocol for a long time, it is for instance possible to analyse factors related to changes in vehicle design. An other example is the data base on whiplash associated disorders causing a sick leave of more than four weeks duration which is hosted by AGU Zurich (Switzerland, http://www.agu.ch). The collection includes cases from the entire country of Switzerland. Due to the large amount of available data, specific topics

concerning technical, medical as well as biomechanical aspects of soft tissue neck injuries can be addressed [e.g. Schmitt et al. 2003a]. Yet other in-depth investigations are made by vehicle manufacturers where specialised teams investigate cases in which vehicles of their own production are involved in order to assess the effectiveness of safety measures and identify needs for improvements. Some of these latter accident data bases also include cases where vehicle damage occurred, but no injury was recorded. Such data are particularly helpful for statistical analysis, as they offer the possibility of well-defined control groups, which are not necessarily available in other types of data bases.

Having recognized that the adequate supply of road accident and injury records is perceived to be important for the selection, implementation and evaluation of road safety measures, several approaches such as for example the European STAIRS project (Standardisation of Accident and Injury Registration Systems, 1997-1999), that are aimed at harmonising accident data collections in order to allow more comprehensive and comparable studies, are under development. Little such efforts are underway for workplace, household or sports injuries which in view of increasing globalization and international mobility may cause, among other, problems with liability and insurance coverage.

2.2 Injury criteria, injury scales and injury risk

Injury criteria are important tools to assess the severity of accidental loading and the risk of sustaining injury thereof. By definition, an injury criterion correlates a function of physical parameters (e.g., acceleration, force) with a probability of a certain body region to be injured in a specific fashion (e.g., concussion, fracture). Injury criteria are generally derived from experimental studies in combination with empirical evidence, and their formulation and validation requires an extensive stepwise extrapolation procedure, since, as mentioned above, experiments on living humans at traumatic levels are excluded.

First, in addition to the concept of "injury criterion", two further criteria have to be introduced, viz., damage criterion and protection criterion. While an injury criterion is intended to describe the property with respect to injury tolerance of living tissue, a damage criterion relates to post mortem test objects as surrogate for the living human. In both cases, a threshold value for the exposure to a quantity calculated from physical parameters is established above which, i.e., if the exposure exceeds the threshold, the test

tissue in question is injured with respect to its anatomical or physiological structure in a specific fashion in more than 50% of all experiments made or accidental exposures under comparable conditions. A protection criterion is obtained when postulating a threshold value on the basis of measurements performed with an anthropomorphic test device (see section 2.5.1) as a human surrogate. In the latter case, the relation to human injury tolerance levels is mainly derived from empirical investigations. It is thereby assumed that a healthy middle-aged adult does on average not sustain injuries of the kind addressed by the particular criterion if he or she is exposed to loading conditions which are comparable to the ones defined in the protection criterion. The actual risk of injury can then be estimated with a risk function which relates the probability to be injured to the criterion developed (i.e. the underlying mechanical properties measured). A threshold value is defined such that given a certain loading scenario, represented by a certain value for the criterion, the risk of sustaining injury does not exceed a percentage of 50%.

However, the definitions of injury, damage, and protection criteria are often not clearly differentiated and thus the term injury criterion is widely used for any index meant to quantify impact or accidental loading severity. Protection criteria, in turn, are determined in standard test procedures, mostly for use in automotive laboratories, which have been defined and established internationally. These procedures are described in section 2.5. In the chapters 3 to 8 specific injury criteria for each body region are presented.

Scales to classify the type of an injury are based on medical diagnosis and were developed for injuries sustained in traffic accidents. The most widely used such scale is the Abbreviated Injury Scale (AIS), which was first developed in 1971 as a system to define the severity of injuries throughout the body and which is regularly revised and up-dated by the Association for the Advancement of Automotive Medicine. The AIS is a standardised system for categorising the type and severity of injuries arising from vehicular crashes (Table 2.1) and is oriented towards the survivability of an injury, i.e., each category represents a certain threat-to-life associated with an injury. Thus, AIS is an anatomically based, global severity scoring system that classifies each injury in every body region by assigning a code which ranges from AIS0 to AIS6. Higher AIS levels indicate an increased threat-to-life. AIS0 means "non-injured" and AIS6 "currently untreatable/maximum injury".

As a result, the AIS severity score is a single, time independent value for each injury and every body region. The severity is described regarding its importance to the whole body, assuming that the described injury occurs to

Table 2.1 The AIS classification.

AIS code	injury
0	non-injured
1	minor
2	moderate
3	serious
4	severe
5	critical
6	untreatable

an otherwise healthy adult. However, it has to be noted that the AIS considers only the injury and not its consequences. Clinical complexity, cost of surgical treatment and long-term sequelae are in particular not taken into account. Hence, severe impairments such as blindness or life-threatening complications due to nosocomial infections occurring in a hospital are not coded as severe injury, because they do not represent an initial threat-to-life.

Moreover, the AIS is not a linear scale in the sense that the difference between AIS1 and AIS2 is comparable to the one between AIS5 and AIS6. It does therefore not make sense to calculate average AIS codes (AIS 3.7, e.g., is a meaningless number). To describe an overall injury severity for one person with multiple injuries, the maximum AIS (MAIS) is used. The MAIS represents the highest AIS code sustained by one person on any part of the body, even if the person in question sustained several injuries of the same severity level at different body parts. If, for example, a car occupant sustained AIS2 injuries on the head and the legs but no injuries classified higher, the MAIS will still be MAIS2.

To account for a better representation of patients with multiple injuries, the Injury Severity Score (ISS) was introduced which is regularly updated like the AIS scale [latest version: AAAM, 2005]. The ISS distinguishes six different body regions: head/neck, face, chest, abdomen, extremities including pelvis, external (i.e. burns, lacerations, abrasions, contusions independent of their location on the body surface). For each of these regions the highest AIS code is determined. Then the ISS is calculated by the sum of the squares of the AIS codes of the three most severely injured

body regions. Thus the minimum ISS is 0 and the maximum ISS is 75 (i.e. three AIS5 injuries). If an AIS6 injury is recorded, the ISS is automatically assigned to 75. ISS values higher than 15 are regarded as major trauma. Several studies have shown that the ISS correlates quite well with several measurement systems such as mortality [e.g. Baker and O'Neill 1976] or long-term impairment [e.g. Campbell et al. 1994].

In addition to the AIS, other scales are used to specify injuries of particular body regions in more detail. The Quebec Task Force [Spitzer et al. 1995], for example, established a scaling scheme to categorise soft tissue neck injuries (see chapter 4). Further scales address impairment, disability and societal loss through ratings of the long-term consequences of the injury by assigning an economic value. An example is the Injury Cost Scale ICS [Zeidler et al. 1989], by which the average costs for an injury is determined taking into account the costs for medical treatment and rehabilitation, loss of income and disability. Further economic scales are the Injury Priority Rating IPR [Carsten and Day 1988] and the HARM concept [Malliaris et al. 1985] applied by the US government.

One of the most crucial problems in trauma-biomechanics is the assessment of the relationship between injury severity and a mechanical load which causes this injury, i.e. to find a relationship that allows assigning probabilities which describe the likeliness that a certain mechanical load (e.g. determined by an injury criterion) will cause a particular injury. This is particularly important because without such correlations, it is rather useless trying to interpret any results obtained, for instance, in crash tests. Hence, it is necessary to perform well equipped laboratory experiments using human surrogates to determine the biomechanical response and corresponding injury tolerance levels and consequently establish so-called injury risk functions.

For the determination of injury risk curves basic statistical methods are applied of which the maximum likelihood method, the cumulative frequency distributions, and the Weibull distribution are most often used. In chapter 3, an example with respect to head injury is presented. For in-depth information however with respect to the application of statistical methods to the often complex and difficult analysis of accident and injury data the reader is referred to statistical text books. Great care has to be exercised in such analyses; among the various problems which may arise when transforming experimental results to (real world) injury risk functions, are

- the small number of tests performed,
- differences in the biomechanical response between the human surrogates used in testing (e.g. cadavers) and living humans,

- differences between the population of the test subjects and the real world population at risk,
- a large spread of data due to different test conditions used by different researchers,
- a large number of possible injury mechanisms and injuries that might occur.

Basically the same limitations apply when using data from accident statistics instead of experimental results to fit injury risk curves. Nonetheless, decades of trauma-biomechanics' research have provided a sufficiently large number of sources that allow establishing a number of well-founded relationships that link mechanical loads to injury probability - at least for certain injuries and injury mechanisms, respectively. However, work is this area is by far not finished and revisions of existing criteria on the basis of new findings are not uncommon.

2.3 Basic technical definitions and accident reconstruction

The reconstruction of accidents is an indispensable procedure in the field of trauma-biomechanics because relations between loading and injury under physiological conditions manifest themselves only in real-life accidents. Likewise, accident reconstructions are often required for forensic purposes likewise in criminal and in civil cases.

The reconstruction of an accident consists of the mathematical analysis of the event in question on the basis of the laws of classical mechanics. Other than laboratory experiments, however, accidents in everyday life occur under largely uncontrolled and unmonitored conditions. Depending on the extent, quality and accuracy of the available documentation, therefore, the specialist in accident reconstruction has to apply assumptions and approximations at quite different levels of complexity. While an accident in a skiing competition may be covered by various video recordings or the traces in a traffic accident may accurately be documented by the police, a fall from a ladder during household activities is hardly documented. All information is of importance in a reconstruction process. Much as in a puzzle, various sources of information have to be combined in order to produce a reliable and conclusive account of the events; this may include as different facts as the sequence of traffic lights in a vehicle-pedestrian impact and the bending stiffness of a pole in case of a sports incident. A scrutiny of the accident scene is always indispensable.

Experience from formerly performed tests under laboratory conditions or the results from well documented "comparable" accidents may furthermore be of help. Of paramount importance is often the collaboration with the medical forensic expert in that injury patterns can provide useful clues for the purpose of accident reconstruction; for example, from the particular appearance with which street dirt presents itself under the skin the direction of a fall can be deduced, etc.

Missing documentation or missing visible evidence may pose problems in accident reconstruction. In case of vehicle collisions, e.g., uncertainties might arise if due to anti-locking systems no skid marks are produced. Furthermore, reconstruction becomes more difficult when no or only marginal vehicle deformation occurs. In order to reduce the repair cost, modern vehicles are designed such that in collisions of low intensity nearly no damage is caused (or at least it is not visible from the outside and therefore often mistaken as not existing by laymen). However, missing visible damage does neither mean that there was no collision at all nor that the energy transmitted might not have been sufficient to cause injury to the occupant.

In what follows, a number of basic mechanical definitions are first reviewed. A distinction has thereby to be made between rigid body mechanics and continuum mechanics (for a comprehensive theory of classical mechanics including the formulations used here, the reader is referred to text books). Both approaches are associated with approximations which have to be carefully assessed in each application, and are widely used in trauma-biomechanics.

Mass, time, position are the fundamental quantities upon which the theory is built.

Rigid body mechanics: mass m, moment of inertia I, time t, position $\vec{r}(t)$, angular velocity $\vec{\omega}(t)$.

The position vector $\vec{r}(t)$ denotes the location of the centre of mass of a rigid body as function of time. Further quantities derived thereof are the velocity of the centre of mass $\vec{v}(t) = \frac{d}{dt}\vec{r}(t)$, furthermore the acceleration $\vec{a}(t) = \frac{d^2}{dt^2}\vec{r}(t)$. The linear motion of the rigid body is described by Newton's second law of motion:

$$m \cdot \vec{a}(t) = \sum_i \vec{F}_i(t) \tag{2.1}$$

whereby the sum extends over all forces $\vec{F}_i(t)$ acting on the body. The

spatial orientation of the body, in turn, is obtained from the angular momentum equation,

$$I \cdot \frac{d}{dt}\vec{\omega}(t) = \sum_i \overrightarrow{M_i}(t) \tag{2.2}$$

with the angular acceleration $\frac{d}{dt}\vec{\omega}(t)$ and the sum over all moments $\overrightarrow{M_i}(t)$ acting on the body. Because of the solidification principle, these equations also hold for deformable bodies, however, the centre of mass is not at a constant location with respect to the contour of the body in such cases.

Continuum mechanics: density $\rho(\vec{r}, t)$, time t, velocity field $\vec{v}(\vec{r}, t)$.

The density $\rho(\vec{r}, t)$ as well as the velocity field $\vec{v}(\vec{r}, t)$ refer to a specific, fixed location $\vec{r}$ in space (this approach is often denoted as Euler representation of the continuum). The equation of motion reads

$$\frac{\partial}{\partial t}(\rho\vec{v}(t)) + (\rho(\vec{v}, \vec{\nabla}))(\rho\vec{v}(t)) = \vec{k}(\vec{r}, t) + (\vec{\nabla}, \hat{\sigma}(\vec{r}, t)) \tag{2.3}$$

where $\vec{k}(\vec{r}, t)$ denotes field forces, e.g., gravity, while the stress tensor $\hat{\sigma}(\vec{r}, t)$ includes contact forces. $\vec{\nabla}$ is the Nabla operator. The angular momentum relation requires that the stress tensor $\hat{\sigma}$ be symmetric. Conservation of mass furthermore yields the continuity equation

$$\frac{\partial}{\partial t}\rho + (\vec{\nabla}, \rho\vec{v}) = 0 \tag{2.4}$$

While positions and deformations are obtained from equations 2.3 and 2.4, the relation between the stress tensor and deformations has to be formulated as constitutive equation which describes the mechanical properties of the continuum. In case of biomechanics, constitutive relations are usually highly nonlinear and involve visco-elasticity and plasticity.

While rigid body models are characterised by a finite degree of freedom associated with a set of ordinary differential equations, in continuum mechanics partial differential equations prevail and the number of degrees of freedom is infinite. For numerical treatment, the partial differential equations have to be approximated in special formulations, of which the Finite Element approximation is most often used in trauma-biomechanics (see chapter 2.6).

Within the framework of a rigid body approximation (equations 2.1, 2.2) for the description of an impact event, empirical investigations and

laboratory experiments have shown that the acceleration of the centre of mass experienced by a body limb under the influence of impact forces is an important parameter to assess the severity of an impact. In many practical cases, the modulus of the acceleration $\vec{a}(t)$ is thereby often related to the acceleration due to gravity, g (1g = 9.81 m/s^2), because we are constantly exposed to gravity such that we can relate a given amount of acceleration with everyday experience. Yet, the acceleration which a body undergoes during the course of an accident varies with time, such that the quantities "peak acceleration" and "mean acceleration" along with the corresponding intervals in time should always be clearly distinguished in order to prevent misunderstandings.

Reconstruction techniques have mostly been developed systematically for traffic accidents. In such cases, a number of specific parameters relating to an involved vehicle have found to be useful for an assessment of the loading situation of occupants.

- The collision or impact velocity of a vehicle is probably the parameter most frequently quoted in the public. In accident reconstruction, the travelling speed or, more accurately, the speed before the beginning of any braking action, is sometimes of importance when investigating whether or under which circumstances a collision could have been avoided.
- The collision-induced velocity change (delta-v) of the vehicle under consideration is, however, in most cases more useful for describing the collision severity when the effects of the collision on the occupants are concerned. The delta-v corresponds approximately to the integral of the translational vehicle deceleration over the collision time for collisions which are characterised by a single impact without significant rotation of the vehicle. Yet, in complex collision situations (roll over, fall over the roadside, etc.) delta-v may not be a well defined parameter.
- The energy equivalent speed (EES) characterises the amount of energy needed to deform a vehicle. In fact, the EES represents the impact velocity into a rigid barrier that would have been necessary to cause the same permanent deformation as observed in the real world accident. The EES is given in [km/h] and can be obtained for many vehicle types from so-called EES catalogues. These catalogues are established on the basis of crash tests conducted under well-defined test conditions.
- A further parameter used to describe impact conditions is the vehicle overlap. This is the extent to which the vehicle and the collision partner (e.g. another vehicle or a barrier in a crash test) overlap. The overlap is generally presented as the percentage of the total width of the vehicle under consideration covered by the opposing vehicle (or wall).

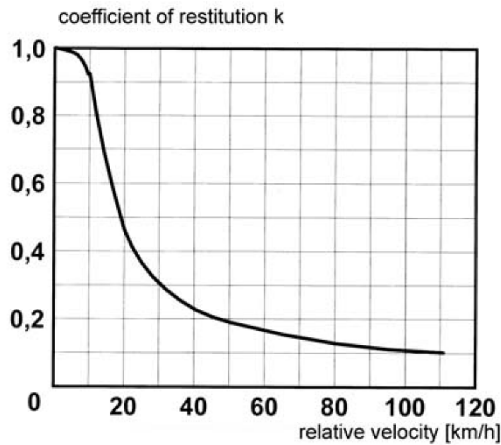



Fig. 2.1 Schematic representation of the relation between the coefficient of restitution and the relative velocity for a frontal impact on a rigid barrier for a passenger car [adapted from Appel et al. 2002].

- From basic mechanics, the principles of elastic and plastic impact and the accompanying coefficient of restitution (k-factor) are used to characterise the elastic and plastic (i.e. permanent) components of the deformation suffered in the impact. Figure 2.1. shows, as an example, the dependency of the coefficient of restitution on the impact velocity (against a rigid wall).

Today, most traffic accident reconstructions are performed with facilitating computer programmes such as Carat [IBB 2002], PC-Crash [DSD 2000] or EDCRASH [EDC 2006] which are thoroughly validated and whose application procedures are well defined. Rigid body dynamics are thereby implemented (equations 2.1, 2.2). Using such programmes, two methods can be distinguished in principle: "forward" and "backward" calculation. In the first case, the kinematics before the collision are assumed, i.e. initial directions of motion, velocities etc. are assigned to the collision partners. Then the actual collision and the final positions of the collision partners after the collision are determined by integration of the rigid body equations whereby tire and collision forces are taken into account. Finally, the positions and traces that were recorded on the actual accident scene are compared with the results of the calculation. In an iterative process, the input parameters are adjusted and the procedure is repeated until a satisfactory match between the results obtained in the calculation and the available accident data is reached. The backward calculation method starts by investigating the final positions of the collision partners. Next, the

motions after the impact are reconciled with the traces found (e.g. skid marks) giving the positions at impact, again utilizing rigid body approximations. Eventually, the initial parameters that lead to the determined course are obtained. Graphics are finally used to give a visual account of the reconstructed accident.

Because of the large mass ratio car occupant/vehicle, the influence of car occupants, likewise of other objects which are not rigidly connected with the vehicle on the colliding vehicle motion can be taken into account in an approximate fashion. This is not the case in motorcycle or bicycle accidents, where the programmes mentioned above can only be applied under restricted conditions and the results have to be interpreted carefully.

Collision phases, not only in traffic accidents, are usually associated with deformation processes for which the application of approximations based on continuum mechanics (equations 2.3 and 2.4 and associated constitutive relations) are required. Because of liability issues mostly, car manufacturers are reluctant to make the Finite Element models which they use to assess the crashworthiness of their vehicles generally available. Various types of simplification are therefore made in general purpose reconstruction programmes. One way is to assume a segmented stiffness distribution of the vehicle's front, and then to integrate the equations of motion of the two vehicles over the collision duration. Another way, often employed in European reconstruction programmes, is to assume the collision duration to be infinitely short (in comparison to the pre- and post-crash motion of the vehicles) and to calculate only the transfer of the (linear and rotational) momentum from one vehicle to the other. The EES values mentioned above may, for both approaches, be used as control values to obtain not only conservation of momentum, but also the energy balance over the collision.

Once a vehicle motion is reconstructed, the motion of the occupants or of an impacted external victim (pedestrian, two-wheeler) during impact can be estimated, again using rigid body models. Furthermore, indications with respect to the occupant loading can be obtained. Further extrapolations, in particular concerning injuries, require however expertise beyond the classical (mechanical) accident reconstruction. The same holds true for accidents occurring at the workplace, household or in sports. Given appropriate circumstances and a careful adaptation to the situation in question, traffic accident reconstruction models and computer programmes can also be utilised in other accidents. For the purpose of injury analysis, the subsequent application of a Finite Element model of the human body may give useful clues.

Finally, accidents are sometimes reconstructed by a one-to-one reproduction on location or in the laboratory with the original installations,

vehicles, sports accessories, etc. This procedure is particularly important in non-traffic related accidents as well as in the course of legal procedures where large claims justifying the often considerable expense of such tests are involved.

2.4 Experimental models

All mechanical characteristics relating to the behaviour in time of the human body, of a part of it, of an organ or tissue when it is subjected to dynamic mechanical loading is subsumed under the term "biomechanical response". The head-neck kinematics as observed in a rugby scrummage or the force-deflection characteristics of the chest due to a frontal vehicle impact are examples for the biomechanical response of the human body. Besides such mechanical changes, the biomechanical response can also lead to physiological changes like neck pain, oedema of the lung or aberrations of the ECG.

A thorough knowledge of the biomechanical response is indispensable for the development of measures for injury prevention and mitigation. Since accident situations as such are highly dynamic by their nature, relevant tests to investigate the biomechanical response of the human body have generally to be conducted under corresponding loading conditions. Nevertheless, whenever extrapolations to dynamic conditions are possible, quasi-static tests are made because of the much simpler installations needed for such tests.

The analysis of the biomechanical response of the human body is not only crucial for an understanding of injury mechanisms, but it is also needed for the definition and verification of injury tolerance thresholds. An important aspect thereby is the biological variability, in particular, age-related changes are prominent. For a reliable measurement of an injury risk function a large amount of experimental data is therefore required. As biological material for testing purposes is not readily available, a careful examination of statistics is of primary importance. Response data may also be restricted by the impossibility to install instrumentation at the desired location, the application of different test protocols by different research teams and a small number of tests. Bearing in mind that many of the relevant studies represent pioneering work in trauma-biomechanics' research dating back to the 1940s, some of these shortcomings can be explained with the lack of adequate measurement instrumentation and the lack of knowledge at that time. In the chapters dealing with the

biomechanical response of the different body regions these problems are discussed in more detail. Furthermore, section 2.5.1 is devoted to the utilisation of human surrogates (dummies) used in impact testing where the response data obtained from the surrogate have to be interpreted in light of biological verisimilitude.

In the following, experimental models used to determine the biomechanical response of the human body are briefly discussed. Five different models can thereby be distinguished, viz., human volunteers, human cadavers, animals, mechanical human surrogates and mathematical models.

Volunteer experiments are, for obvious reasons, restricted to the low severity range only, i.e. well below any level thought to be possibly injurious. The pain threshold is often taken as the upper limit up to which mechanical loads are applied. Advantages related to volunteer tests are first of all the use of the "correct" anatomy and physiological state. Moreover, the influence of the muscle tone can be studied and effects like the bracing prior to a collision can be considered. However, the cohorts used for volunteer tests are usually not statistically representative for the population at risk. Particularly, females, children and the elderly are strongly under-represented in volunteer data available. Difficulties also arise with the instrumentation as load cells can often not be brought to the location of interest (e.g. the head's centre of gravity or the first thoracic vertebra), even a rigid external fixation is difficult to reach because of the skin. Advances in high-speed video camera technology along with sophisticated mathematical treatment have considerably contributed to the improvements of such results. Ciné-radiography has sometimes been used to monitor the response of the skeleton to impact, e.g. by Ono and Kaneoka 1997 to investigate the motion of cervical spine vertebrae. As the number of subjects tested in this fashion is particularly small, questions of scaling to other groups of humans as well as to other (more severe) impact severity are all the more critical.

Human cadavers (usually denoted as post mortem human subjects (PMHS) or post mortem test objects (PMTO)) are the second type of model used to determine human biomechanical response. Despite the great anatomical similarity to the living human (a PMTO may to some extent be compared with a sleeping human), several influencing factors have to be considered. First, the age of the PMHS is often high. Age-related degeneration is therefore often prevalent in the cadaver cohort available for a test series, for example, in case of osteoporosis, fracture is observed too frequently. Second, the lack of pressure in the lungs and the blood vessels, the absence of muscle tone, as well as differences due to preparation

techniques used (i.e. embalmed vs. non-embalmed cadavers) significantly influence the biomechanical response. Fresh cadavers, however, were shown to be good models for the detection of fractures, vessel ruptures and lacerations [Kramer 1998/2006]. Nonetheless, physiological responses (e.g. the neck pain or ECG aberrations) cannot be addressed with such models. For the investigation of the response of a single body part only, for instance of the leg (see chapter 7), isolated cadaver parts are used. Here the connection to the rest of the body has to be mimicked in the test set-up in an appropriate way.

Animal models have a limited significance for human trauma-biomechanics. Nevertheless, anaesthetised animals offer the only possibility to investigate physiological reactions to severe mechanical loading. Animal experiments also allow a comparison between living and dead tissue and thus give important input to the proper interpretation of cadaver tests. However, due to differences in anatomy and physiology, the possibilities of scaling the results obtained, particularly with respect to injury thresholds, are limited.

Further models used in trauma-biomechanics include mechanical human surrogates, i.e. anthropomorphic test devices (ATDs) as well as mathematical (computational) models. Because of their importance (e.g., all regulations on vehicle occupant safety are formulated in terms of measurements made on ATDs), these models are discussed in separate sections in the following.

The objective of impact testing in the laboratory consists of a realistic simulation of accident scenarios and of the determination of the mechanical loading that a human victim possibly sustains in such an accident. Most laboratory test set-ups are thereby made for vehicle crash testing mostly because of the comprehensive regulatory coverage of vehicle safety. In the automotive industry, extensive usage of crash facilities is made for the assessment of restraint systems as well as for the development of new measures in passive safety to reduce the number and severity of injuries sustained in automotive accidents. Yet, laboratory tests are also used to certify football helmets or ski bindings, etc.

Real world accident scenarios are manifold. Thus, only selected impact conditions which are thought to be of relevance are simulated in crash testing. Bearing in mind the need of repeatability and comparability of test results along with the cost and time related to crash testing, several standards were developed that define the exact test protocols, the evaluation process, as well as the protection criteria to be derived thereof. In section 2.5 such standardised test procedures are described in detail.

Three different categories of automotive crash tests can be distinguished,

viz., full scale tests, sled tests, and component tests (Figure 2.2). The basic principles with respect to laboratory practise, evaluation of results and documentation thereby also apply to non-automotive testing and certification procedures as regards, for example, the closing force of elevator doors or the strength of nets used by the fire brigade.

In full scale impact tests, a vehicle impacts an obstacle or another vehicle

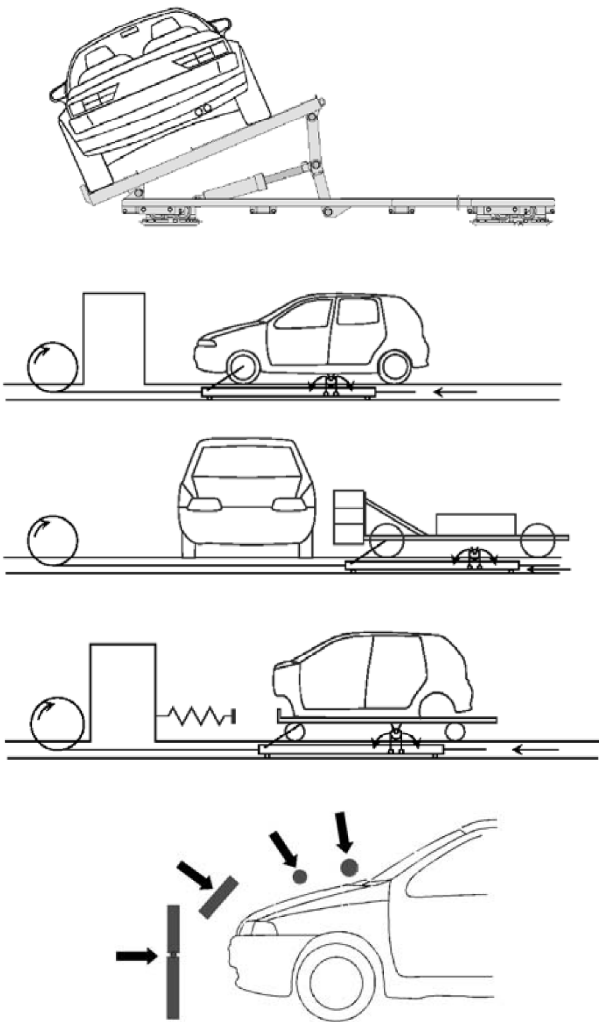



Fig. 2.2 Different methods of crash testing. From top to bottom: full scale testing (rollover test, frontal and lateral impact), sled testing and different impactors used in pedestrian safety testing of the front of a car.

or is impacted by a moveable object (e.g. a barrier as used in side impact tests). Anthropomorphic test devices (i.e. crash test dummies) represent occupants located in the vehicle under consideration, and the kinematics and the mechanical loading of the dummy is recorded during impact. Full scale crash tests have the advantage that the actual vehicle properties, e.g. the deformation characteristics, are inherent in the results. These properties influence the acceleration response of the vehicle and consequently the loading of the occupants. In addition to passive safety issues, full scale tests provide information about the repair costs to be faced after a collision and are therefore also performed by insurance companies with respect to the rating of the insurance premium. Full scale tests are also used for non-biomechanical purposes, e.g. to check the fuel system integrity or the braking system.

While in full scale tests the interaction between the restraint systems and the deformation characteristics is investigated, sled tests are primarily used to analyse the isolated behaviour of restraint systems or vehicle components (e.g. a driver's seat). For this purpose, parts of the vehicle or the components of interest are mounted on a sled. The sled is accelerated or decelerated, respectively, in a controlled manner without damaging the test rig. Consequently, the sled including parts of the assembly can be re-used, thereby significantly reducing the associated cost. The disadvantages of this type of test are, among other, the restriction that the vehicle loading may only be unidirectional, and that the vehicle acceleration pulse must be established by a prior full-scale test or, in prototyping, by e.g. computer simulations.

Component tests form a third type of testing. Here, in quasi-static as well as dynamic tests various aspects concerning single parts of the car body may be investigated. In quasi-static tension tests, for instance, the strength of the seat belt attachment points is examined. Furthermore, using devices such as the free motion head form (FMH) the compliance and the energy dissipation properties of the vehicle interior are assessed. The FMH is a head form mounted on a propelling device such that it can be projected towards onto the vehicle structure in question under different angles. Using other dummy parts (e.g. lower and upper limb surrogates and head forms simulating children and adult heads), pedestrian safety is assessed by evaluating the deformation properties of the vehicle front.

2.5 Standardised test procedures

All new car models must pass numerous tests related to occupant safety before they may be brought into circulation. These tests are thereby partly different in the various countries around the world; the most important however being the EU and the US. In Europe, the corresponding procedures are laid down in the regulations of the UN Economic Commission for Europe (ECE). ECE R94, for example, describes the test procedure for frontal impact protection, while in ECE R95 the side impact test is defined. Recently, these regulations have been incorporated in the EC directives, where 96/27/EC contains ECE R95 and 96/79/EC includes ECE R94, for example. For the sake of simplicity, we refer to the older ECE Rxx designation in the following chapters. In the United States, the Federal Motor Vehicle Safety Standards (FMVSS) are incorporated in the Federal Register 49 CFR part 571. Since most car makers aim to sell their cars on a global market, the differing safety standards in different parts of the world constitute a considerable problem. International harmonisation of tests and the international recognition of test results obtained in a certified laboratory are important aspects in worldwide trade. To this end, numerous bilateral trade agreements between countries, furthermore free trade initiatives, UN, US and EU ("Cassis de Dijon" principle) activities were made or are under way. Therefore, the UN/ECE/WP.29 has been designated to develop harmonised regulations, called GTR (Global Technical Regulations).

In addition, instruments, machines, installations, sports accessories, etc. which are in daily use are subjected to a myriad of regulations, guidelines and recommendations made by government bodies, manufacturers, insurance companies, sports associations and consumer organisations. In different countries, quite different regulations and practises can be found. A general overview can hardly be made, in each individual case the internet has to be consulted. International trade however requires increasingly a mutual recognition of standards.

As can be seen in Tables 2.3 and 2.4, the ECE regulations and the FMVSS are quite similar and include many corresponding regulations. However, differences arise for the types of dummies requested, the test conditions prescribed or the evaluation of the tests (Figure 2.9). Furthermore, different threshold values for occupant loading apply in some cases. The requirements stated in both the ECE regulations and FMVSS are also often adapted in other countries and therefore can be considered the most powerful safety regulations worldwide. For complete and up-to-date information the reader is advised to check the corresponding internet sites.

Table 2.3 ECE regulations (for details see http://www.unece.org).

regulation	collision type	impact velocity	test conditions	comments
R94	frontal	56 km/h	40% overlap, deformable barrier	2 Hybrid III dummies
R12	frontal	48..53 km/h	rigid wall	concerning deformation of the steering assembly
R33	frontal	48..53 km/h	rigid wall	concerning stability of passenger compartment
R12	frontal	24 km/h	impactor test	determining force on body block impactor
R95	side	50 km/h	moveable, deformable barrier, 90° angle	1 EuroSID at driver position
R32..34	rear-end	35..38 km/h	moveable, rigid barrier (mass: 1100 kg)	integrity of the petrol system
R42	minor collisions	2.5, 4 km/h	pendulum	checking safety in operation only
R44	child restraint systems (CRS)	50 km/h	sled tests	different dummies used depending on CRS
R16	seats	-	static	recliner moment, deformation
R17	seats	-	sled tests, 20 g	seat anchorage to vehicle body, head restraint geometry
R14	belts	-	static	e.g. deformation

Table 2.4 FMVSS regulations (for details see http://www.nhtsa.dot.gov).

regulation	collision type	impact velocity	test conditions	comments
571.208 (latest version phase 2)	frontal	25 mph	100% overlap, 0 - 30° rigid barrier	2 unbelted Hybrid III dummies (50% male)
		35 mph	100% overlap, 0° rigid barrier	2 belted Hybrid III dummies (50% male)
		25 mph	100% overlap, 0° rigid barrier (max. 5° oblique)	2 unbelted Hybrid III dummies (5% female)
		35 mph	100% overlap, 0° rigid barrier (max. 5° oblique)	2 belted Hybrid III dummies (5% female)
		25 mph	40% overlap, 0° deformable barrier	2 belted Hybrid III dummies (5% female)
		-	various configurations, firing of airbags	various dummies in OOP situations
571.204	frontal	30 mph	100% overlap, rigid barrier	steering assembly rearward displacement
571.212	frontal	30 mph	100% overlap, rigid barrier	concerning the mounting of the windscreen
571.203	frontal	15 mph	impactor test	determining force on body block impactor

Table 2.4 ctd. FMVSS regulations (for details see http://www.nhtsa.dot.gov).

regulation	collision type	impact velocity	test conditions	comments
571.203	frontal	15 mph	impactor test	determining force on body block impactor
571.214	side	33.5 mph	moveable, deformable barrier, oblique impact	2 SID dummies used
571.301+ 303	rear-end, front, side	30 mph	moveable, rigid barrier (mass: 1800 kg)	fuel system integrity
581	minor collisions	2.5 mph (rear), 5 mph (front)	pendulum/barrier	checking safety in operation only
571.213	child restraint systems (CRS)	30 mph	sled tests	different dummies used depending on CRS
571.210	seats	-	static tests	e.g. deformation
571.209	seat belts	-	static tests	e.g. deformation

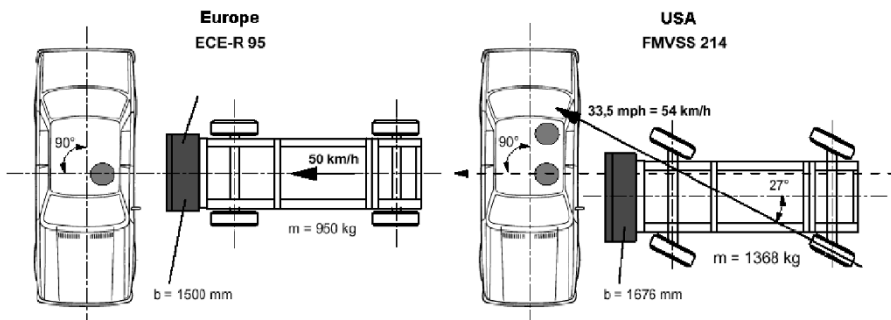



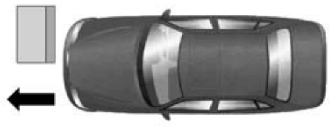
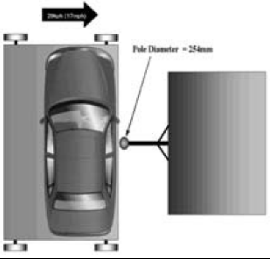
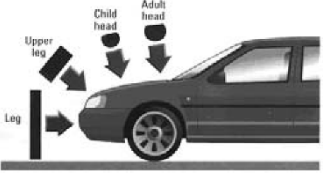
Fig. 2.3 Different test conditions for lateral impact are required by the ECE (left) and the FMVSS (right).

Table 2.5 Frontal impact threshold values.

	FMVSS 208	ECE R94
dummies	Hybrid III 50% male, 5% female	2 Hybrid III 50% male
head	HIC 15 < 700	HPC < 1000 a3ms< 80 g
neck	Nij <= 1.0, {-4.17kN < Fz < 4.0kN} (Hybrid III 50% male) {-2.62 kN < Fz < 2.52 kN} (Hybrid III 5% female)	Mext<57 Nm
thorax	a3ms <= 60 g, deflection < = 63 mm (Hybrid III 50% male.)/ deflection <= 52 mm (Hybrid III 5% female)	deflection < 50 mm VC < 1.0
femur	axial force < 10 kN	not exceeding defined force corridor
knee		deflection < 15 mm
tibia		axial force < 8 kN TI < =1.3

Table 2.6 Side impact threshold values.

	FMVSS 214	ECE R95
dummies	ES-2, SIDIIs	1 EuroSID
head	HIC 36 < 1000 (both dummy types)	HPC < 1000
thorax	A max < 82 g (both dummy types) d max < 42 mm (ES-2)	VC < 1.0
abdomen	F < 2.5 kN (ES-2)	internal force < 2.5 kN
pelvis	F < 5.1 kN (SIDIIs) / F < 6 kN (ES-2)	pubic force < 6 kN

Tables 2.5 and 2.6 summarise the requirements for occupant protection as defined in ECE R94 and FMVSS 208 for frontal impact and in ECE R95 and FMVSS 214 for lateral impact.

More details on the protection criteria mentioned and their threshold values are given in chapters 3 to 8 for the respective parts of the human body. It has to be noted that neither the ECE nor the FMVSS include regulations concerning occupant safety in rear-end collisions although they occur frequently and cause enormous health problems. To fill this gap, a new test procedure was developed by AGU Zurich in collaboration with Autoliv GmbH Germany, GDV Munich and the University of Graz [Muser et al. 1999]. A modified version of this procedure has been incorporated into an ISO standard by ISO/TC22/SC10.

In addition to the crash tests required by governmental regulations, consumer tests are performed. As legislation provides a minimum statutory standard of safety for new cars only, and because the results from the governmental tests are not necessarily published, it is the aim of consumer tests to encourage car manufacturers to exceed these minimum requirements and make the results of these consumer tests publicly available. Thus, consumers can obtain reliable and accurate comparative information regarding the safety performance of individual car models.

In Europe, dummies have been used in consumer tests to determine the occupant loading even before regulations demanded such tests. Thus, the public was made aware of the importance of passive safety issues. Moreover, consumer tests are characterised by ranking systems which are intended to give the consumers the possibility to assess and compare the occupant protection potential of different vehicle types. Such rating schemes often include dummy symbols with coloured body regions ranging from green (i.e. low loading) to red, and final star ratings where the number of stars correlates with the total number of credits gained in the assessment. This total number of credits cannot only be obtained from vehicle performance during the crash tests, but credits are also given for safety features concerning prevention or driver education (e.g. an acoustic "fasten seat belt" reminder will influence the final grade positively in EuroNCAP tests).

To date, the most important consumer tests are the so-called New Car Assessment Programs (NCAP). NCAP testing is performed in Europe (EuroNCAP), Australia, Japan and the US. The test conditions and ranking systems differ for different NCAP agencies. Table 2.7 lists the tests performed by EuroNCAP. It should be noted that in the US and e.g. Australia, NCAP tests were performed by government agencies long before non-government institutions like EuroNCAP came into play.

Table 2.7 Test conditions applied by the Euro-NCAP [http://www.euroncap.com].

impact	test conditions	
frontal impact	64 km/h, deformable barrier, 40% overlap	
side impact	50 km/h, Trolley fitted with a deformable front is towed into the driver's side of the car	
pole test (head protection)	29 km/h, car is propelled sideways into a rigid pole	
pedestrian impact	40 km/h impactor speed, various impacts on front structure	

2.5.1 Anthropomorphic test devices

Standardised tests required the usage of well defined and validated test objects. An anthropomorphic test device (ATD) is a mechanical model of the human body that is used as a human surrogate in crash testing. ATDs are in particular designed such that mechanical loading parameters can be measured at impact levels which would be injurious for a living human. To this end, a dummy is made of steel or aluminium (e.g. skeleton), polymers (joints, skin) and foam (flesh) and is equipped with several accelerometers and load cells to record acceleration, force or deformation. To date various

types of ATDs - commonly called crash test dummies - are available whereas each ATD is designed for one specific type of impact only.

In automotive engineering, ATDs are used in the homologation tests required for new vehicles, and in safety device testing to evaluate the occupant protection potential. To a somewhat smaller extent, dummies are also used in the aircraft industry for similar purposes. Historically, the first dummies were developed for the use in aviation, to test parachutes and ejection seats.

Test devices and especially devices embodied in official regulations are expected to fulfil a given set of requirements:

- Anthropometry and biofidelity. An ATD should on the one hand represent a human in terms of size, mass, mass distribution, moments of inertia and (sitting) posture and on the other hand display a human-like biomechanical response to impact. The 50th percentile adult male of which the underlying anthropometric data were established in the 1960s from the US population (standing height: 1.751 m, total weight: 78.2 kg) is the most commonly used dummy in automotive crash testing. Other dummy types include the 5th percentile female (h: 1.510 m, w: 49.1 kg) and the 95th percentile male (h: 1.873 m, w: 101.2 kg). 3, 6 and 10 year old child dummies are furthermore available. The biofidelity is assessed on the basis of cadaver and volunteer studies.
- Instrumentation. The crash test dummy should be sensitive to and allow the measurement of parameters that are related to the injury or the injury mechanism to be examined.
- Repeatability and durability. It should be borne in mind that a dummy must continue to record data for later evaluation even if a critical threshold is exceeded during the test, i.e. it should not or only rarely be damaged.

Repeatability (performing the same test repeatedly with the same dummy) and reproducibility (comparing results obtained under the same test conditions with different dummies) require that an ATD be calibrated regularly. Moreover, practical considerations play an important role in dummy design. Dummies should be robust enough to withstand a high number of tests (even with overload) and they should allow easy handling (up to 102 kg!) and adjustment of the posture.

Currently, over 20 different dummy types are available of which not all are, however, included in government regulations. Table 2.8 gives an overview of available ATDs.

The Hybrid III family of dummies consists of a 3-year-old, 6-year-old, 10-year-old, small adult female (5th percentile), mid-sized adult male (50th

Table 2.8 Dummies available and their field of application.

application	anthropomorphic test devices
frontal impact	Hybrid III family, THOR
lateral impact	EuroSID, EuroSID2, SID, SID-HIII, SID IIs, BioSID, WorldSID
rear-end impact	BioRID, RID2
pedestrian	POLAR
children	P0, P3/4, P3, P6, P10, Q-dummies, CRABI
belt	TNO-10
impactor	free motion head impactor, head/hip impactor for pedestrian impact

percentile) and large adult male (95th percentile). These dummies are designed for use in frontal impact tests. The Hybrid III 50th percentile male dummy (Figure 2.4) is the most widely used crash test dummy for the evaluation of automotive restraint systems in frontal crash testing. The dummy is defined in the US Federal Motor Vehicle Safety Standards (FMVSS, contained in the US Federal Register) as well as in the European directives. The skull and skull cap of the Hybrid III 50th percentile male dummy are made of cast aluminium parts with removable vinyl skins. The neck is a segmented rubber and aluminium construction with a centre cable. It accurately simulates the human dynamic moment/rotation flexion and extension response in situations involving high neck loading. The rib cage, in turn, is represented by six high-strength steel ribs with polymer based damping material to simulate human chest force-deflection characteristics. Each rib unit comprises left and right anatomical ribs in one continuous part which is open at the sternum and anchored to the back of the thoracic spine. A sternum assembly connects to the front of the ribs and includes a slider for the chest deflection rotary potentiometer. The angle between the neck and upper torso is determined by the construction of the neck bracket, in which a six-axis lower neck transducer can be incorporated. A two-piece aluminium clavicle and clavicle link assemblies have cast integral scapulae to interface with shoulder belts. A curved cylindrical rubber lumbar spine mount provides human-like slouch of a seated person and mounts to the pelvis through an optional three axis lumbar load cell. The pelvis is made of

a vinyl skin/urethane foam moulded over an aluminium casting in the seated position. The ball-jointed femur attachments carry bump stops to reproduce the upper leg to hip moment/rotation characteristics. While the femur, tibia and ankle can be instrumented to predict bone fracture, the knee is designed to evaluate tibia to femur ligament injury. The foot and ankle simulates heel compression and ankle range of motion.

A further frontal impact dummy called THOR (Test device for Human Occupant Restraint) (Figure 2.5) was developed in recent years. This dummy is also based on the anthropometry of the 50th percentile male. Compared to the design of the Hybrid III, all dummy components were improved except the arms, which are identical with those of the Hybrid III. The facial region of the dummy is, for example, instrumented with unidirectional load cells to asses the probability of facial skull fracture. Furthermore, the biofidelity and geometry of the rib cage was enhanced by the use of elliptical ribs and by improving instrumentation such that the dynamic three-dimensional compression of the rib cage can be determined at four distinct points. A new abdominal assembly was developed such that belt intrusion and compressive displacement at the upper abdomen that might possibly result from an airbag can directly be measured. Changes to the pelvis and the lower limbs increased the sensing capabilities and in addition, the ankle joint was rendered more human-like.

The first side impact dummy (SID) was developed in the late 1970s at the University of Michigan. SID is based on the predecessor of the Hybrid

Fig. 2.4 50th percentile male Hybrid III dummy [Denton ATD Inc.].

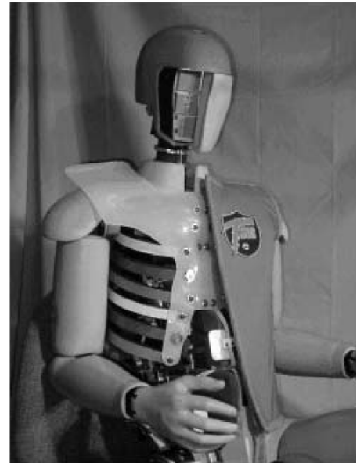



Fig. 2.5 The THOR dummy [Gesac Inc.]

III (the Hybrid II) with an adapted thorax, but without arms and shoulder structures. SID is also sized corresponding to the 50th percentile male and is used in US government-required side-impact testing of new cars (FMVSS 214). The dummy primarily allows a measurement of the injury risk to the head, chest and pelvis. To account for a better head-neck biofidelity, a SID dummy equipped with a Hybrid III head and neck is available (called SID-HIII). It is applied in compliance testing of side-impact head airbags. Additionally, the SID II, i.e. a side-impact dummy representing a 5th percentile female, became commercially available in 2000.

European lateral impact regulations (ECE R95) require the use of the Euro-SID1, the European side impact dummy. In Australian and Japanese impact regulations the Euro-SID1 is likewise prescribed. An updated version, today also accepted for homologation testing, is denoted as ES-2. The original Euro-SID, which was finalised in 1989 represents a 50th percentile adult male. Euro-SID basically consists of a metal and plastic skeleton, covered by flesh-simulating materials. The sitting height is 0.904 m. The total body mass is 72 kg. The dummy which has no lower arms is shown in Figure 2.6. While the head and the legs are that of the Hybrid III, the thorax was developed to analyse lateral impact. Three separate identical ribs covered with flesh-simulating foam are attached to a rigid steel spine box through a system consisting of a piston/cylinder assembly, springs and a damper (Figure 2.6). A special shoulder construction allows the arms to rotate realistically and expose the ribs to direct impacts. The dummy can be

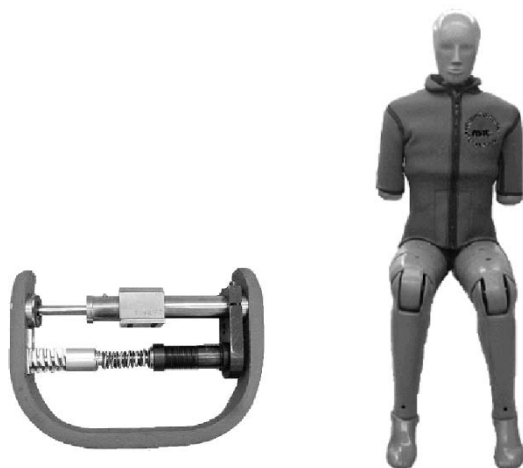



Fig. 2.6 Euro-SID and one of its spring-damper-elements used in the rib cage [Denton ATD Inc.].

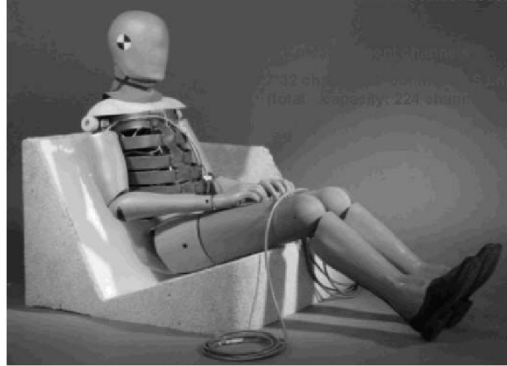



Fig. 2.7 The World-SID [ISO World SID Task Group].

used for side impacts from its left- as well as from its right-hand side.

Further developments in side impact dummies include the Biofidelic Side Impact Test Dummy (BioSID) intended to improve the performance of the current US standard SID series. Although available since 1990, the BioSID was not yet included in FMVSS 214. BioSID has more sensors and a more biofidelic body than SID/Hybrid III, such that it allows the measurement of the thoracic, abdominal and pelvic injury potential as well as the rib deflection and other compression-based injury criteria. By rotating the upper torso by 180 degrees, the dummy can be converted from a left side to a right side impact dummy.

As the automotive industry becomes more global, a harmonised side-impact dummy, denoted as World-SID was developed by a worldwide consortium under the umbrella of the International Standardisation Organisation (ISO). In a comprehensive approach, a mid-sized male side impact dummy for improved assessment of injury risk to car occupants in lateral collisions was thereby developed within the framework of the World-SID programme. Besides an improved biofidelity [e.g. Damm et al. 2006], the World-SID is intended to lead to a worldwide harmonisation in safety regulations and will in the future probably be incorporated in the Global Technical Regulation initiative (GTR) which was created to this end. A first prototype as well as 11 pre-production dummies were evaluated in various laboratories worldwide. The production design of the World-SID was released in March 2004 (Figure 2.7).

So far only dummies for frontal and lateral impact were presented. This is not surprising, as current occupant safety regulations are restricted to these impact directions. Because the assessment of occupant protection in other than frontal and exactly lateral, in particular rear-end impact crash

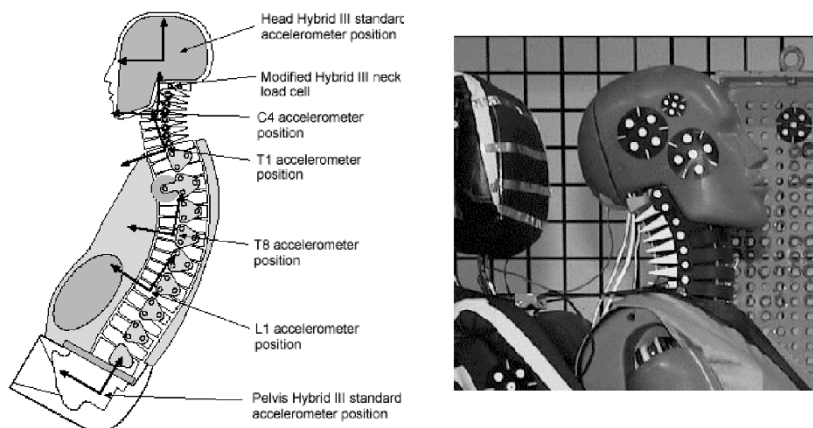



Fig. 2.8 The BioRID makes use of a fully segmented spine [Denton ATD Inc].

tests is not required, there was no need to develop suitable test devices. However, since injuries sustained in rear-end collisions especially neck injuries sustained in low-speed rear-end collisions constitute a major problem in road traffic (see chapter 4), the need emerged to develop anthropomorphic test devices that allow the investigation of these impact conditions.

To date, two different dummies for posterior impact are available, the BioRID and the RID2. Both are mid-sized male dummies have been developed in Europe for assessing the risk of "whiplash" injuries in low-speed rear-end impacts. The main feature of the biofidelic rear-end dummy (BioRID) is its fully segmented spine consisting of 24 segments. Each human spinal pivot point is reproduced. Due to such a detailed representation, a biofidelic spinal movement is observed (Figure 2.8). The rear impact dummy (RID2), in turn, is based on the THOR frontal impact dummy. However, several modifications were made of which the new design of the neck, which consists of seven aluminium discs, and of the flexible thoracic and lumbar spine are most relevant in view of an the analysis of the neck injury risk. Both the RID2 and the BioRID were developed and validated for pure rear-end impacts with a movement of the spine exclusively in the anterior-posterior plane. More recently, an improved neck for the RID2, called RID3D, was presented, allowing also oblique rear-end and even low speed frontal impacts to be analysed. Although these dummies offer the possibility for better investigation of the head-neck kinematics, difficulties in handling arise due to the increased flexibility of the spine. The seating procedure, for example, is a quite

intricate task compared to a Hybrid III.

In addition to the dummies described above, several specially designed test devices exist. These test devices are generally used for one particular test purpose only.

- The TNO-10 dummy is a loading device for testing vehicle safety belts in a frontal crash situation. The dummy represents a 50th percentile male adult with respect to size and weight distribution. For reasons of simplicity the dummy has no lower arms and only one lower leg assembly combining the two human legs.
- The Child Restraint Air Bag Interaction dummy (CRABI) is used to evaluate air bag exposure to infants restrained in child safety seats that are placed in the front seat. CRABI dummies come in three sizes: six-month-old, 12-month-old and 18-month-old. Further child dummies like the Q-dummies, or infant dummies representing the newborn (P0) and the nine-month-old (P3/4) are available in addition to the child dummies of the Hybrid III family.
- The POLAR dummy (current version: POLAR II) has been designed to simulate more accurately the kinematics of the human body during car-pedestrian collisions. Standing 175 cm tall and weighing 75kg, the new dummy will help to gather more accurate data on injuries sustained by pedestrians.
- Test devices representing only parts of a dummy are used. The free motion head form (FMH) models a human adult head. Mounted on a propelling device, parts of the vehicle interior may be subjected to a simulated head impact. These tests are required by some safety regulations, e.g. FMVSS 201. Other impactors are used to test the behaviour of a car front with respect to pedestrian safety. These impactors, representing an adult head, a child head, an upper leg and a lower leg are, for example, used in the EC directives and the Euro-NCAP (New Car Assessment Programme) test scheme (see section 2.5).
- A 50th percentile torso-shaped body block which is solely used to test the deformation characteristics of the steering assembly, is required for testing in e.g. ECE R12. Parts of ECE R12 have, however, been superseded under certain conditions by ECE R94 and are therefore not required any more in Europe.

2.6 Numerical methods

Thanks to the continuous advancements in computer technology as well as in numerical methods, mathematical modelling has become gradually more detailed and more powerful. Today, computer simulations are an important tool in trauma-biomechanics and are applied in all areas of safety engineering such as vehicle crashworthiness design and accident reconstruction; in addition, computer models are successfully used in human body modelling, thereby addressing in particular biomechanical response and possible injury mechanisms.

The most widely used simulation techniques are the multi body system (MBS) approach based on rigid body dynamics (equations 2.1, 2.2) and the finite element (FE) method, a particular formulation of continuum mechanics (equations 2.3, 2.4). Multi body systems are sometimes also referred to as lumped mass models in that complex structures such as a human organ or a vehicle are condensed into one or more rigid units connected by mass-less elements like springs and dampers (see e.g. the Lobdell thorax model, chapter 5). Besides, the solidification principle of basic mechanics as well as St.Venant's principle of continuum mechanics are always in the background. Multi body systems and FE representations of subunits are furthermore often combined. Likewise, a multi body system can contain flexible subunits, e.g., a cantilever or a plate which can be approximated with models having only few degrees of freedom.

In a multi body system the various elements are connected by kinematic joints. The presence of the kinematic joints restricts the relative motion between adjacent bodies and hence reduces the degrees of freedom of the system. Different types of joints are available, for example translational, revolute and spherical joints, of which each is characterised by a specific number of degrees of freedom. Additional kinematic constraints (e.g. spring/damper elements) can be applied. The rigid bodies themselves are characterised by their inertial properties and by the location of the above mentioned joints only. For the modelling of contact interactions (e.g. head-windscreen impact) and for visualisation purposes, geometrical shapes may be associated with rigid bodies. For the modelling of human body or dummy parts, ellipsoids are often used. Other geometrical primitives include planes and cylinders.

The behaviour of the MBS system is analysed by subjecting the system to external forces such as an acceleration field corresponding to a crash pulse or the forces associated with a fall from a window. This technique has proved its strength especially in whole body response modelling.

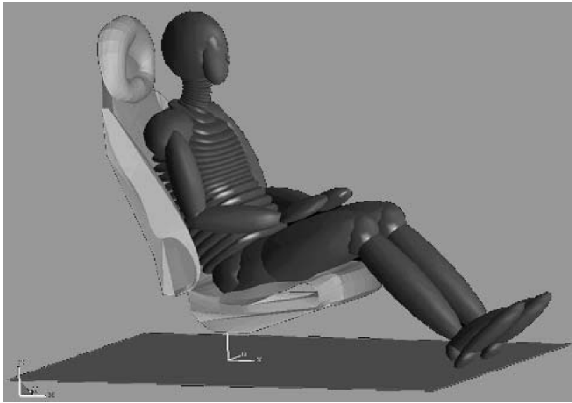



Fig. 2.9 MBS model showing a BioRID dummy seated [adapted from Schmitt et al. 2003b].

Approximating the human body by various rigid bodies that are linked by joints and by assigning inertia and mass properties to those bodies, the gross human body kinematics during impact can be simulated. First models were presented already in the 1970s. To date, a wide range of validated models is available. In particular, dummies are well suited to be modelled as a MBS, because the geometrical and mechanical properties (inertia, mass, joint properties) of the dummy components are clearly defined. Figure 2.9 illustrates an example of a MBS that includes a model of a BioRID. The example was established using the software MADYMO [TNO 2001] which is probably the most frequently used MBS programme for occupant safety problems.

In the finite element (FE) method, originally derived from Galerkin's theorem, a continuous system is reduced to a discrete numerical model consisting of well defined elements (e.g. hexahedrons, quadrilaterals, bars). Each element consists of a fixed number of nodes. The degree of freedom of the whole FE model is therefore restricted by the number of nodes. Depending on the boundary conditions applied and the geometry of the mesh, in particular for those elements that share common nodes, the degree of freedom of the whole FE model is given. A detailed description of the finite element method can be found, among other, in Bathe (1996) and Zienkiewicz (1994). However, it should be noted that the nature of the problems to be solved in trauma biomechanics (e.g. non-linear material behaviour, large deformations in short time intervals) require specialised approaches to the solution of the models. In general, FE programmes used in this field (PAM-CRASH [ESI 2001], LS-DYNA [Livermore 1999], or

Radioss [Mecalog 2000]) are based on explicit time integration formulations. These formulations are based on the differential equations of motion of the nodes rather than on the equilibrium of inertial, field and contact forces (implicit formulation). This approach requires less calculation effort and easily lends itself for implementation in parallel computers. On the other hand, more care must be taken to control the numerical stability than in implicit formulations.

The FE method offers the possibility of detailed analysis of the response to impact of both the vehicle and the human body (Figure 2.10). For example, regarding the response of the head and brain, FE models offer the possibility to investigate the stress distribution in the brain during impact. Such results are important with respect to the understanding of diffuse brain injuries (see chapter 3), but can hardly be addressed in experiments. Other complex biomechanical phenomena, for instance the influence of muscle activity or the interaction of fluid flow and the changing geometry of the surrounding tissue, can be approached by finite elements [e.g. Schmitt et al. 2002].

In summary, both the MBS and the FE technique offer their specific advantages and disadvantages in the field of crash simulations. The FE method allows for detailed studies of complex geometries and problems concerned with contact interactions. With respect to crash simulations, the study of local deformations and stress distributions are important

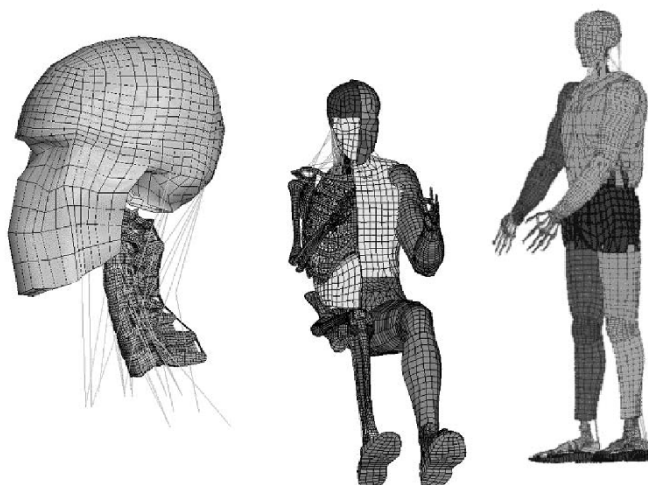



Fig. 2.10 The FE technique used in human body modelling. A detailed model of the head-neck complex on the left [adapted from Schmitt et al. 2002] and a whole body model on the right [adapted from Iwamoto et al. 2002].

advantages of this method. As such, this method can also be used for the analysis of injury mechanisms by modelling a specific part of the human body. However, a detailed representation of a complex geometry leads to an enormous number of elements and therefore a large number of unknowns to be calculated. In case of nonlinear constitutive properties of the involved materials as well as large deformations, the enormous computational cost often associated with the FE method represents a major limitation. Parallel processing may thereby be of help. To date, large computer systems are able to handle FE models with millions of unknowns (e.g. about 700’000 elements for simulations of compatibility tests with two cars modelled in detail), with computation times of several days. In contrast, its capability to represent complex kinematic connections efficiently makes the MBS approach particularly attractive. Additionally, computation times required are generally much shorter than for FE calculations since usually only a comparably small number of ordinary differential equations, though mostly stiff, are to be treated. Hence the MBSs are widely used as design tools as they are well suited for optimisation studies involving many design parameters.

With respect to human body modelling, general problems arise that both techniques have to cope with. The choice of parameters to describe the material behaviour of the living human tissue requires the availability of experimental data with respect to the deformation characteristics of living tissues. Such data are hardly available, and, if yes, often associated with a large uncertainty because of general biological variability on the one hand, and limitations of the particular experimental procedure chosen for the constitutive tests on the other. Furthermore, the validation of human body models, especially those intended for use in several different impact conditions, is crucial but remains a complex task.

To conclude, both methodologies can be reasonably used in the field of general impact and injury analysis. Depending on the purpose, either the best suited technique has to be chosen, or a combination of both methods can be considered. Such an integrated (or hybrid) approach is for instance realised in simulations of interactions of a car occupant and a deploying air bag. In this case an FE model is used to model the airbag while the human is represented by a MBS. Various other studies are presented where a MBS is used to model the gross motion while FE models are included for detailed analysis of single structures, for example, an ice hockey player crashing into a rink board. As of today, numerical models are included at basically all stages of the development process of safety devices.

Despite the widespread and rapidly increasing use of simulation techniques and their potential to reduce the number (and associated cost) of

crash tests, numerical simulations are not yet included in the general vehicle safety standards. This can partly be attributed to the fact that general guidelines for simulations and especially for quality control are not existing, but would be required if crash simulations were embodied in safety regulations.

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3 Head Injuries

Head injury sustained in accidents continues to be a leading cause of death and disability even though considerable advancement in the understanding of head injury mechanisms and the introduction of different measures to prevent such injury (e.g. helmets, vehicle restraint systems) has resulted in the reduction of the number and severity of head injuries.

In this chapter, a brief review of the head anatomy is followed by the description of possible head injuries and the underlying injury mechanisms. Further, the biomechanical response of the head as investigated in various experimental studies and injury criteria that were derived thereof to quantify the impact response of the head in crash testing are discussed. Aspects of head injuries in sports are considered in an own section. Finally, principles of head injury protection are presented.

3.1 Anatomy of the head

The human head (cranium) can be regarded as a multi-layered structure with the scalp being the outermost layer followed by the skull, the meninges and eventually the central nervous system that represents the innermost tissue.

The scalp is about 5 mm to 7 mm thick and consists of the hair-bearing skin, a subcutaneous connective tissue layer, and a muscle and fascial layer. Applying a traction force to skin of the head, these layers move together as one. Below the scalp there is a loose connective tissue and the periosteum (i.e. a fibrous membrane) that covers the bony skull.

The adult skull is a complex structure consisting of several bones fused together and associated suture lines (Figure 3.1). The only facial bone connected to the skull through free moveable joints is the mandible. Thickness and curvature of the bones can vary substantially.
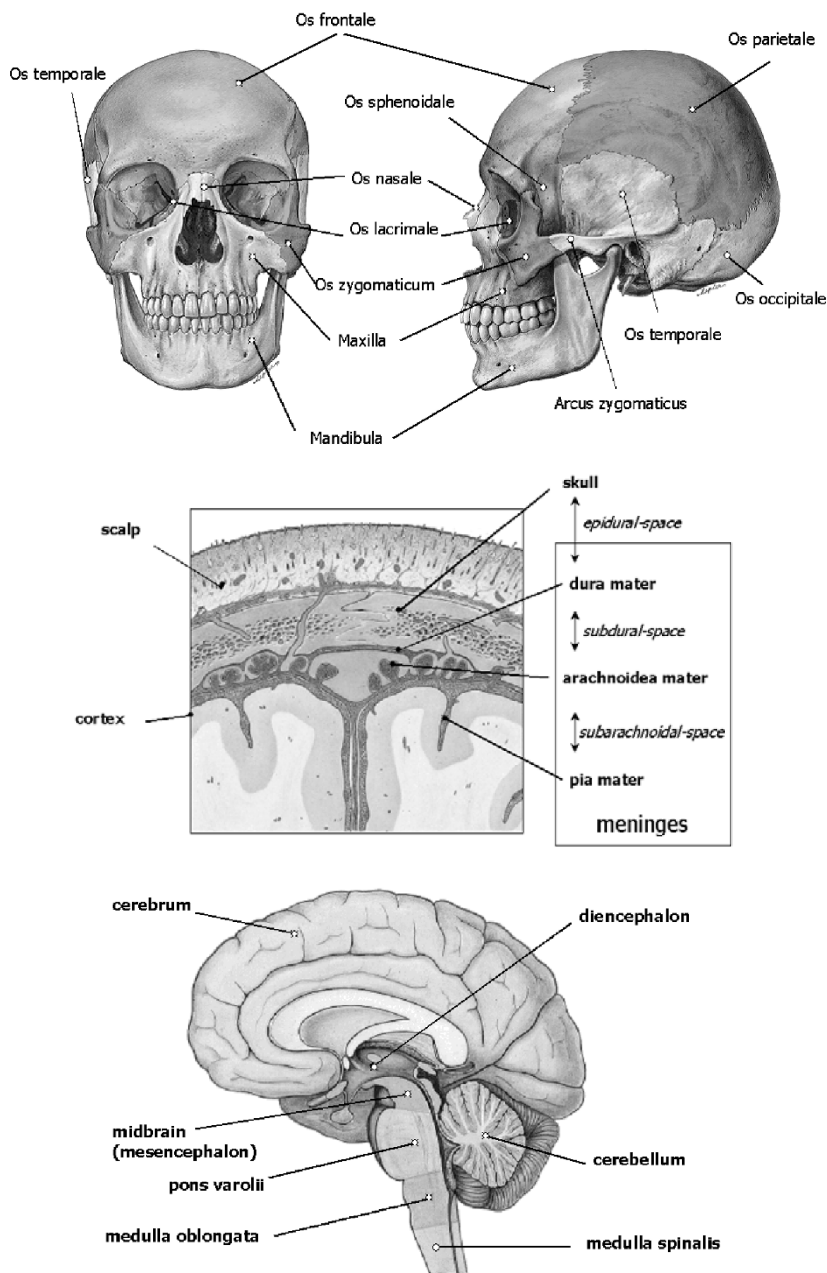
Fig. 3.1 Anatomy of the head: bony structures of the skull (top), the meninges (middle), and the brain (bottom) [adapted from Sobotta 1997].

The inner surface of the cranial vault is concave with an irregular plate of bone forming the base. This base plate contains several small holes for arteries, veins and nerves as well as a large hole (foramen magnum) through which the brainstem passes into the spinal cord.

Three membranes called the meninges protect and support the spinal cord and the brain and separate them from the surrounding bones (Figure 3.1). From outside to inside, we find the dura mater, the arachnoidea mater, and the pia mater. The dura mater is a tough, fibrous membrane, while the arachnoidea mater resembles a spider-web. Both membranes are separated by a narrow space, the subdural space. Analogously, the subarachnoidal space separates the arachnoidea mater and the pia mater. The pia mater covers the surface of the brain, dipping well into its fissures. Cerebrospinal fluid (CSF) fills the subarachnoidal space and the ventricles of the brain and thus cushions the brain (and the spinal cord) from mechanical shock. As CSF constantly circulates and surrounds the brain on all sides, it serves as a buffer and helps to support the brain's weight.

Several blood vessels cross the meninges supplying the brain and the scalp. The so-called bridging veins, i.e. the veins that bridge the subdural space, are of particular interest as they may be subject to injury through tearing (see section 3.2.).

Finally, the central nervous system consisting of the brain and the spinal cord is located at the centre of the head. Structurally and functionally the brain can be divided into five parts: cerebrum, cerebellum, midbrain, pons and medulla oblongata (Figure 3.1).

3.2 Injuries and injury mechanisms

Most important injuries to the head are those to the skull and the brain including the meninges. Figure 3.2 gives an schematic overview on possible head injuries. In principle, head injuries are characterised as open or closed depending on whether the dura mater is injured (open) or not (closed). Soft tissue injuries to the scalp and face commonly occur in automotive accidents. The resulting injuries include contusion and laceration but are generally regarded to be of minor importance. Likewise facial injuries, to the eyes or ears for example are considered minor injuries and therefore are mainly rated as AIS1 or AIS2. These injuries will not be discussed here.

More severe head injuries can arise from fractures. Facial fractures include fracture of the nasal bone, which occurs most frequently, and
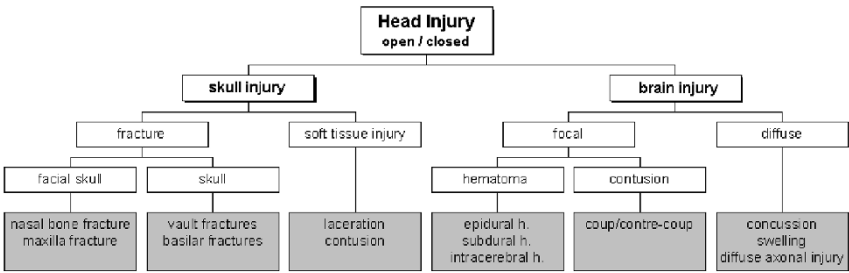
Fig. 3.2 Possible injuries to the head.
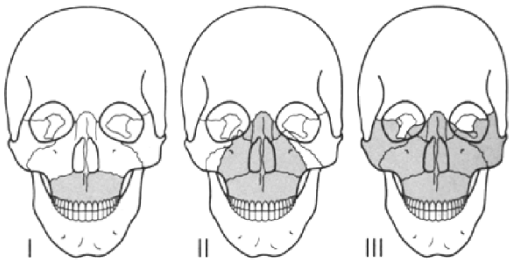
Fig. 3.3 Three types of facial fractures as classified by LeFort [adapted from Vetter 2000].

maxillary fractures. The latter are considered serious with AIS grades of up to 3. Figure 3.3. shows the LeFort classification that is used to categorise maxillary fractures. Examples of head injuries classified according to the AIS scale are presented in Table 3.1.

With respect to the skull, fractures are divided into basilar and vault fractures (i.e. all other fractures not occurring at the basis of the skull). As for basilar fractures it should be noted that those are still not that easy to visualise radio-graphically so that diagnosis can be difficult.

Injuries to the brain are clinically classified into two broad categories: diffuse injuries and focal injuries. Diffuse brain injuries form a spectrum ranging from mild concussion to diffuse white matter injuries. The most common form of such brain injury is mild concussion (fully reversible, no loss of consciousness). Particularly in sports, mild traumatic brain injury (MTBI) is often diagnosed (see section 3.5). A more severe form of concussion is cerebral concussion which is characterised by immediate loss of consciousness. The outcome of patients suffering from cerebral concussion strongly depends on whether there are associated brain injuries or not [Melvin and Lighthall 2002]. Diffuse axonal injury (DAI) describes disruption to the axons in the cerebral hemispheres and the subcortical

Table 3.1 AIS classified head injury [AAAM 2005].

AIS code	description
1	skin/scalp: abrasion, superficial laceration face: nose fracture
2	skin: major avulsion vault fracture: simple, undisplaced mandible fracture: open, displaced maxilla fracture: LeFort I and II
3	basilar fracture maxilla fracture: LeFort III total scalp loss single contusion cerebellum
4	vault fracture: complex, open with torn, exposed or loss of brain tissue small epidural or subdural hematoma
5	major penetrating injury (>2cm) brain stem compression large epidural or subdural hematoma diffuse axonal injury (DAI)
6	massive destruction of both cranium and brain (crush injury)

white matter. Brain swelling may be superimposed on diffuse brain injuries, adding to the effects of the primary injury by increased intracranial pressure.

Focal brain injuries are lesions where the damage is locally well-defined. Possible focal injuries are hematoma and contusions. Contusion is the most frequently found lesion following head impact. Generally, contusion occurs at the site of impact (coup contusion) and at site opposite the impact (contre-coup contusion). Contre-coup contusions are considered more significant than coup-contusions [Melvin and Lighthall 2002]. As for hematoma, three different types are distinguished depending on the site of the bleeding: epidural hematoma, subdural hematoma and intracerebral hematoma (Figure 3.4). Epidural hematoma, i.e. bleeding above the dura mater, is observed as a result of trauma to the skull and the underlying meningeal vessels. It is therefore not due to brain injury. Usually skull

Fig. 3.4 Bleeding into the epidural space is called an epidural hematoma and can cause brain contusion [adapted from Vetter 2000].
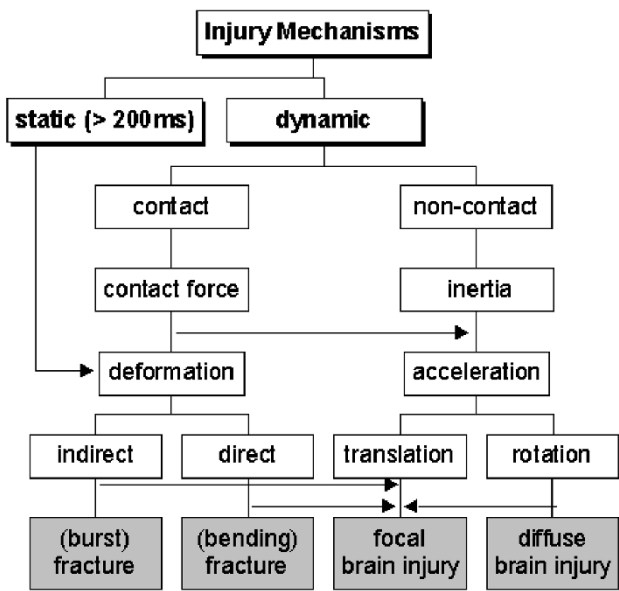
Fig. 3.5 Possible mechanisms for head injury.

fracture is associated, but an epidural hematoma may also occur in the absence of fracture. If the hematoma is found below the dura mater, it is called a subdural hematoma. Three sources were identified for subdural hematoma: lacerations of cortical veins and arteries by penetrating wounds, large-contusion bleeding into the subdural space, and tearing of bridging veins between the brain's surface and the dural sinuses. The mortality rate of this type of hematoma exceeds 30% in most studies [Melvin and Lighthall 2002]. Intracerebral hematomae are well-defined homogeneous collections of blood within the brain and can be distinguished from
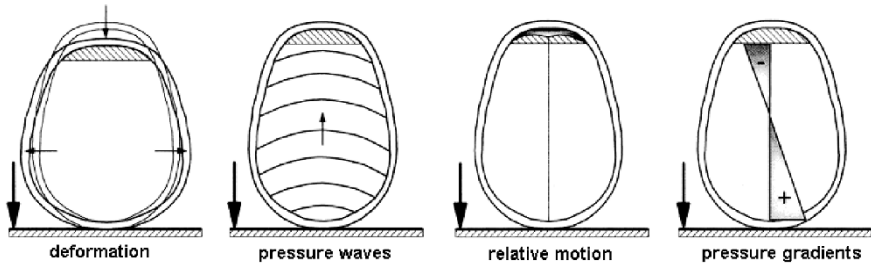
Fig. 3.6 Different injury mechanisms for contact impact; fractures do not necessarily occur [adapted from Vetter 2000].

contusions by e.g. computer tomography.

The mechanisms causing head injuries are manifold. In principle, injuries can result from static and dynamic loading (Figure 3.5). For our purpose, static loading is defined as a load lasting for more than 200 ms. Under such static loading the head deforms until it reaches a maximum deformation. Then the skull fractures, often leading to multiple fractures. In automotive accidents, however, this type of loading is rare. Dynamic loading is the predominately loading scenario. Two types, contact and non-contact loading, are distinguished, each resulting in a different head response. Direct contact of the head to (or from) an object can cause the skull to deform, possibly resulting in direct fractures (mostly due to bending and often close to the impact location) or in indirect fractures (burst fractures oriented in the direction of the force vector). Furthermore, after deformation of the head local brain injury (even without fractures) like epidural hematoma or contusion as well as scalp injuries are observed. Additionally, rapid contact loading produces stress waves that propagate in the skull or the brain (Figure 3.6). Wave propagation in the brain may lead to a pressure gradient with positive pressure at the site of impact (coup) and negative pressure on the opposite side of the impact (contre-coup). Such a mechanism is proposed for the generation of intracranial compression which causes focal injuries of the brain tissue and bruising. However, it is not yet fully understood whether the injury is due to negative pressure (tensile loading) or due to a cavitation phenomenon [Viano 2001]. In addition, the pressure gradient can give rise to shear strains within the deep structures of the brain.

Contact loading may also result in a relative motion of the brain surface with respect to the inner surface of the skull base. Surface contusions on the brain (so called gliding contusions) and tearing of the bridging veins (causing subdural hematoma) can be the consequences.

In non-contact situations, the head is loaded exclusively due to intertial forces, i.e. acceleration (or deceleration) of the head. Acceleration (or deceleration) of the head results in inertial loading. Acceleration can either be translational or rotational. Translational acceleration generally results in focal brain injury while rotational acceleration also causes diffuse brain injury. As an exception, subdural hematoma, i.e. a focal skull injury, may arise due to acceleration induced relative motion between brain and skull tearing the bridging veins. Furthermore, acceleration response of the head does, of course, also occur in contact loading. Thus the mechanisms described above apply in the same way.

It should be noted that headaches, which are often erroneously thought to be due to a "head injury", caused either by contact or non-contact mechanisms, may also be initiated by lesions in the upper area of the cervical spine. Therefore, a comprehensive analysis of the occupant dynamics must be undertaken in order to prevent premature diagnoses of cerebral concussion or mild traumatic brain injury (MTBI).

3.3 Mechanical response of the head

Many cadaver studies on head impact have been carried out to investigate the mechanical response properties of the head. In general, the impact responses were described in terms of head acceleration and impact force and therefore depend on the inertial properties of the head and surface impacted. For a 50th percentile male, the average head mass is 4.54 kg and the average mass moments of inertia are $I_{xx} = 22.0 \times 10^{-3}$ kgm^2, $I_{yy} = 24.2 \times 10^{-3}$ kgm^2, and $I_{zz} = 15.9 \times 10^{-3}$ kgm^2 [e.g. Beier et al. 1980].

Table 3.2 Peak force for fracture at different regions of the skull.

impact area	force [kN]	reference
frontal	4.2	Nahum et al. 1968
	5.5	Hodgson et al. 1971
	4.0	Schneider and Nahum 1972
	6.2	Advani et al. 1975
	4.7	Allsop et al. 1988
lateral	3.6	Nahum et al. 1968
	2.0	Schneider and Nahum 1972
	5.2	Allsop et al. 1991
occipital	12.5	Advani et al. 1982

In these cadaveric studies, mainly drop tests against a rigid flat surface were performed. Table 3.2 summarises the peak force values reported for fracture at different sites of the head. Furthermore the acceleration response of the head was investigated. When measuring the acceleration of the head, two problems arise: firstly, accelerometers cannot be mounted at the centre of gravity of the head and secondly, the head is not a rigid body. Therefore several methods for measuring the acceleration have been proposed [e.g. Padgaonka et al. 1975]. It is also recommended to measure the head rotational acceleration so that the acceleration of the head's centre of gravity can be computed thereof. Nonetheless there remain some uncertainties as the exact stiffness distribution of the skull is generally not known.

As a result of extensive cadaver tests focusing on head acceleration, the Wayne State University Cerebral Concussion Tolerance Curve, abbreviated as the *Wayne State Tolerance Curve (WSTC)*, was established [Gurdjian et al. 1953, Lissner et al. 1960, Gurdjian et al. 1966]. The WSTC indicates a relationship between the duration and the average anteroposterior translational acceleration level of the pulse that accounts for similar head injury severity in head contact impact (Figure 3.7). Clinically observed prevalence of concomitant concussion in skull fracture cases was used to relate cadaver impacts to brain injury. In fact, 80% of all concussion cases also had linear skull fractures [Melvin and Lighthall 2002]. Gurdjian and colleagues assumed that by measuring the tolerance of the skull to fracture loads, one is effectively inferring the tolerance to brain injury.
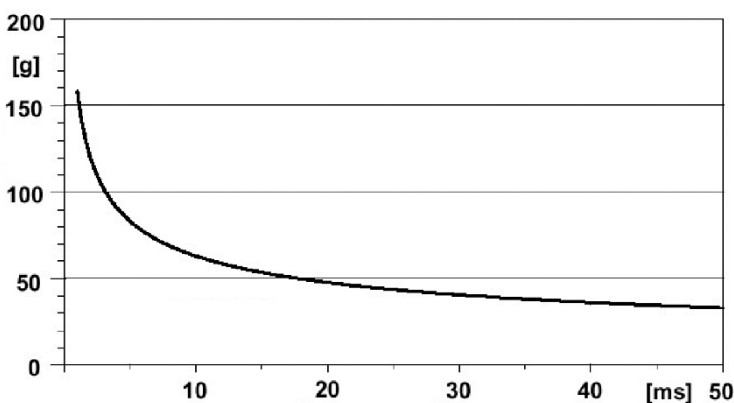
Fig. 3.7 The Wayne State Tolerance Curve (acceleration vs. duration of acceleration pulse) [adapted from Krabbel 1997].

Table 3.3 Test conditions of the experiments the WSTC is based upon.

pulse duration	test objects	test set-up	response measured	injury criterion
2 - 6 ms	cadavers	drop test	acceleration at the back of the head	skull fracture
6- 20 ms	cadavers and animals	impact test	acceleration of skull, brain pressure	pathological changes
> 20 ms	volunteers	sled tests	whole body acceleration without head impact	concussion, state of consciousness

Combinations of acceleration level and pulse duration that lie above the curve are thought to exceed the human tolerance, i.e. they cause severe, irreversible brain injury. Combinations below the curve do not exceed human tolerance, but may result in reversible injury. As the original WSTC covers a time duration range of 6 ms only, the curve was extended for durations longer than 6 ms using animal and volunteer data. Figure 3.7 presents the modified curve; the test conditions used to obtain the data are given in Table 3.3. As can be seen, the head can withstand higher acceleration for shorter durations.

The WSTC is supported by experiments conducted in Japan which led to the Japan Head Tolerance Curve (JHTC) [Ono et al. 1980]. JHTC was mainly obtained from experiments with primates and scaling of results to humans. Differences between the WSTC an JHTC are negligible for time intervals up to 10 ms, and only minor differences exist for longer durations.

When the WSTC is plotted in a logarithmic scale, it becomes a straight line with a slope of -2.5. Based on this finding, Gadd et al. (1961) proposed a first head injury criterion, the severity index (SI). A modified form of this criterion is still in use today (see section 3.4.1).

Using the WSTC or any criterion developed thereof, restrictions that arise from the test conditions have to be considered. The paucity of data points, the position of the accelerometer (back of the head), the fact that the rotational acceleration is not considered, and the techniques used to scale the animal data are, for instance, major limitations. However, from a biomechanical point of view the main criticism concerns the correspondence of skull fracture and brain injury that was assumed. This hypothesis remains to be verified, as there was no direct demonstration of functional brain damage in an experiment in which biomechanical parameters sufficient to determine a failure mechanism in the tissue were

measured [Melvin and Lighthall 2002].

Bearing in mind that the WSTC is based on direct frontal impact tests, the results can, strictly speaking, not be applied to non-contact loading conditions and to other impact directions, respectively. Nonetheless, WSTC is still the most important data source with respect to the linear acceleration response of the head.

Further experimental studies addressed rotational acceleration which cause diffuse brain injury and subdural hematoma. Besides volunteers and cadavers, primates were subjected to head rotation, where acceleration was measured and the resulting degree of injury was assessed [e.g. Ommaya et al. 1967, Hirsch et al. 1968, Gennarelli et al. 1972]. It was found that the angular acceleration and the according injury thresholds are related to the mass of the brain. Thus, the tolerance limit for the human was obtained by scaling the results from the primate tests (Figure 3.8). Table 3.4 gives tolerance values that are commonly used. However, additional studies on volunteers suggest that much higher tolerance values up to 25000 rad/s^2 may be possible for short durations [Tarriere 1987].

In this section several experimental studies were presented that aimed at predicting head injury from one specific input parameter, i.e. the translational or the rotational acceleration, respectively. However, in the vast majority of head impact situations it can be expected that both translational and rotational accelerations are present and combine to cause brain injury. Accordingly, comprehensive brain injury prediction requires taking into account the various responses of the brain tissue for any combination of mechanical loading. The development of sophisticated
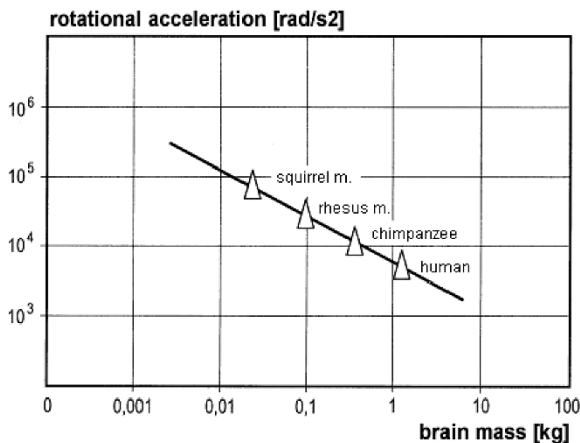
Fig. 3.8 Results from experiments and scaling addressing tolerance towards rotational acceleration [adapted from Krabbel 1997].

Table 3.4 Tolerance thresholds for rotational acceleration and velocity of the brain.

tolerance threshold	type of brain injury	reference
50% probability: $\ddot{\alpha} = 1800$ rad/s^2 for $t < 20$ ms $\dot{\alpha} = 30$ rad/s for $t \geq 20$ ms	cerebral concussion	Ommaya et al. (1967)
$\ddot{\alpha} < 4500$ rad/s^2 and/or $\dot{\alpha} < 70$ rad/s	rupture of bridging vein	Löwenhielm (1975)
$2000 < \ddot{\alpha} < 3000$ rad/s^2	brain surface shearing	Advani et al. (1982)
$\dot{\alpha} < 30$ rad/s: safe : $\ddot{\alpha} < 4500$ rad/s^2 AIS 5: $\ddot{\alpha} > 4500$ rad/s^2 $\dot{\alpha} > 30$ rad/s: AIS 2: $\ddot{\alpha} = 1700$ rad/s^2 AIS 3: $\ddot{\alpha} = 3000$ rad/s^2 AIS 4: $\ddot{\alpha} = 3900$ rad/s^2 AIS 5: $\ddot{\alpha} = 4500$ rad/s^2	(general)	Ommaya (1984)

mathematical models of the head using, for example, the finite element method addresses this task and aims at determining measures for prediction of the head's mechanical response to impact. When combined with results of detailed investigation of the response of the living human, such models promise to contribute substantially to todays understanding of head injury mechanisms and the impact tolerance of the head.

3.4 Injury criteria for head injuries

Although great progress in passive safety, such as the introduction of advanced restraint systems, have been made in the last couple of years to reduce the number and severity of head injuries, there is only one injury criteria in wide use, the Head Injury Criterion (HIC), which was developed more than thirty years ago. Besides the HIC and its European equivalent, the Head Protection Criterion (HPC), the "3 ms criterion" and the Generalised Acceleration Model for Brain Injury Threshold (GAMBIT) are

presented. However, it should be noted that all these criteria are based on acceleration response only. Consequently, injuries that are related to impact force rather than acceleration are not addressed by these criteria. In other words, those criteria do not allow an evaluation of the injury risk of sustaining fractures of the bony structures of the head. The only dummy capable of measuring a force response to facial impact is the THOR dummy (see chapter 2.5.1), but this dummy is not included in recent crash test standards.

3.4.1 Head Injury Criterion (HIC)

The Head Injury Criterion has a historical basis in the work of Gadd (1961), who used the Wayne State Tolerance Curve (WSTC) (see section 3.3) to develop the so-called severity index (SI). In 1971, Versace (1971) proposed a version of the HIC as a measure of average acceleration that correlates with the WSTC. The actual version of HIC was then proposed by the US National Highway Traffic Safety Administration (NHTSA) and is included in FMVSS No. 208. HIC is computed based on the following expression:

$$HIC = max\left[\frac{1}{t_2 - t_2}\int_{t_1}^{t_2} a(t)dt\right]^{2,5}(t_2 - t_1) \tag{3.1}$$

where t_2 and t_1 are any two arbitrary time points during the acceleration pulse. Acceleration is measured in multiples of the acceleration of gravity [g] and time is measured in seconds. The resultant acceleration is used for the calculation. FMVSS 208 requires t_2 and t_1 not to lay more than 36 ms apart (thus called HIC_{36}) and the maximum HIC_{36} not to exceed a value of 1000 for the 50th percentile male. In 1998 NHTSA also introduced the HIC_{15}, i.e. the HIC evaluated over a time interval of 15 ms [Kleinberger et al. 1998]. As for the according threshold value, a maximum of 700 was suggested for the 50th percentile male.

To determine the relationship between HIC and injuries of the skull and brain, available test data was analysed statistically by fitting normal, log normal, and two-parameter Weibull cumulative distributions to the data set, using the Maximum Likelihood method to achieve the best fit for each function [Hertz 1993]. The best fit of the data was achieved with the log normal curve (Figure 3.9).

The probability of skull fracture (AIS ≥ 2) is given by the formula

$$p(fracture) = N\left(\frac{\ln(HIC - \mu)}{\sigma}\right) \tag{3.2}$$

where N( ) is the cumulative normal distribution, $\mu = 6.96352$ and $\sigma = 0.84664$.

Since the data used to establish this risk analysis consists of short duration impacts of typically less than 12 milliseconds, the HIC curve is applicable to both HIC_{15} and HIC_{36}. Thus, the probability of skull fracture (AIS ≥ 2) associated with a HIC_{15} threshold value of 700 for a mid-sized male is 31% and for a limit of 1000 for HIC_{36} (50th percentile male) it is approximately 48 %.

Basically the limitations as described for the WSTC itself apply (see section 3.3). Not taking into account rotational acceleration is often criticised. A further drawback is the lack of a functional relationship between human head injury and the acceleration response measured in an anthropomorphic test device. Despite these limitations, HIC is still the most commonly used criterion for head injury in automotive research.
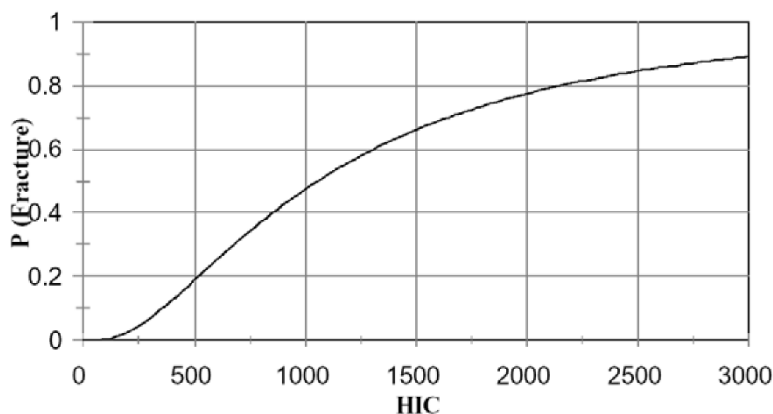
Fig. 3.9 Probability of skull fracture (AIS ≥ 2) in relation to the HIC as determined by Hertz (1993).

3.4.2 Head Protection Criterion (HPC)

The determination of the Head Performance Criterion HPC is required in regulations ECE R94 and R95. Hence, the HPC is used to quantify head impact in both frontal and lateral impact. The definition and the calculation procedure to obtain the HPC are identical to the HIC_{36}. Thus, the corresponding maximum time interval is 36 ms. The according threshold level for frontal and lateral direction is 1000.

If no head contact occurs, the HPC is fulfilled regardless of the acceleration level reached. If the beginning of the head contact can be

determined satisfactorily, t_1 and t_2 (cf. equation 3.1) are the two time points which define a period between the beginning of the head contact and the end of the recording, at which the HPC is at its maximum.

3.4.3 3 ms criterion (a_{3ms})

The "3 ms criterion" is also based on the WSTC. It is defined as the acceleration level exceeded for a duration of 3 ms and should not exceed 80 g [Got et al. 1978]. This criterion is also incorporated in ECE R21 and R25, the regulations dealing with impact of the occupant to interior structures of a vehicle and the impact to the head restraints, respectively. The analogous US regulation, FMVSS 201, as well as the frontal impact regulation FMVSS 208 also require fulfilment of this criterion.

Furthermore, a modification of the a_{3ms} criterion is used in helmet testing. Choosing a duration of 5 ms, the acceleration level shall be less than or equal to 150 g. ECE R22 describes the details of this so-called a_{5ms} criterion.

3.4.4 Generalized Acceleration Model for Brain Injury Threshold (GAMBIT)

In an attempt to combine translational and rotational acceleration, Newman (1986) proposed the *Generalized Acceleration Model for Brain Injury Threshold*, abbreviated GAMBIT. Assuming that a combined load case of translational and rotational accelerations can cause head injury, the following relationship was proposed:

$$GAMBIT = \left[\left(\frac{a(t)}{a_c}\right)^n + \left(\frac{\ddot{\varphi}(t)}{\ddot{\varphi}_c}\right)^m\right]^{\frac{1}{k}} \tag{3.3}$$

Here $a(t)$ and $\ddot{\varphi}(t)$ denote the translational and rotational acceleration, respectively. a_c and $\ddot{\varphi}_c$ represent critical tolerance levels for those accelerations and n, m and k are constants. Fitting the constants by means of statistical analysis and computer simulations to available data, Kramer (1998/2006) presents a solution which reads

$$GAMBIT = \left[\left(\frac{a(t)}{250}\right)^{2,5} + \left(\frac{\ddot{\varphi}(t)}{25}\right)^{2,5}\right]^{\frac{1}{2,5}} \tag{3.4}$$

with $a(t)$ and $\ddot{\varphi}(t)$ given in [g] and [krad/s^2]. Figure 3.10 shows curves of constant GAMBIT obtained by using equation 3.4. The curve for a

GAMBIT of 1.0 was determined to represent a probability of 50% for irreversible head injury. Non-contact head impact accounted for GAMBIT values below 0.62.

Assuming that translational and rotational accelerations contribute equally to the probability of head injury and assuming that the tolerances derived in experiments with either translational or rotational acceleration are also valid in a combined loading scenario, equation 3.3 was simplified to

$$GAMBIT = \frac{a_m}{250} + \frac{\ddot{\varphi}_m}{10} \tag{3.5}$$

with a_m [g] and $\ddot{\varphi}_m$ [krad/s^2] being the mean translation and mean rotational acceleration, respectively, and considering 250 g the maximum tolerable translational acceleration and taking 10 krad/s^2 for the limit for rotational acceleration [Newman 1986]. Thus, GAMBIT of 1.0 represents the overall tolerance value.

To date the GAMBIT still lacks validation and is therefore hardly ever used, nor is it included in any regulations so far.
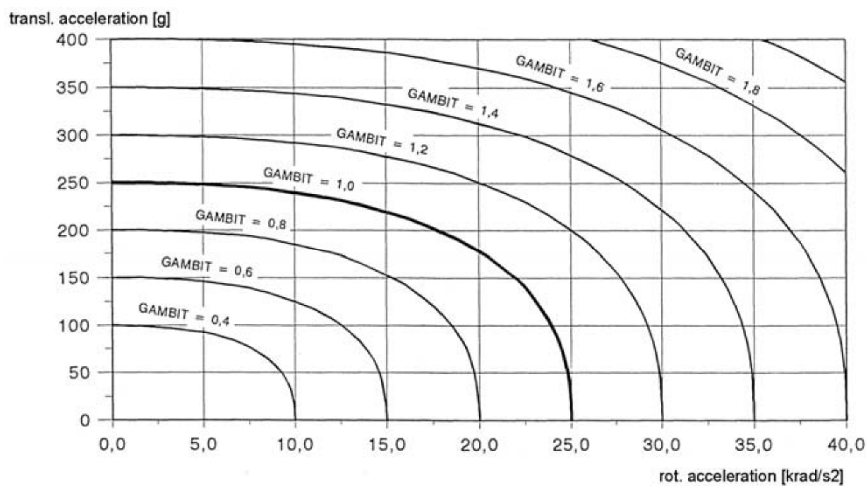
Fig. 3.10 GAMBIT curves for constant GAMBIT values [adapted from Kramer 1998].

3.5 Head injuries in sports

The incidence rate for head injuries depends strongly on the type of sports. Upon team sports at the Olympic Games 2004, Junge et al. (2006) found that 24% of all injuries reported were head injuries. Mild concussion was sustained most often (11%) followed by lacerations (4%), fractures (2%) and contusions (2%). Handball accounted for 42% of the head injuries, soccer for 20%, basketball and hockey for 13%. For other sports, incidence rates of the same order of magnitude were found: skiing/snowboarding 3-15% [e.g. Hunter 1999, Levy et al. 2002], ice hockey 4-18% [McIntosh et al. 2005], baseball 11% head only (28% facial injuries) [Yen et al. 2000], equestrian sports 19%, boxing 16% for concussion [Zarzyn et al. 2003].

Although concussions reported in sports are often classified as minor or mild, they are, particularly in professional sports, a major concern. Repeatedly sustained concussion may result in degeneration of brain tissue. Thus it must be ensured that a player returning to play after a (mild) concussion is fully recovered (see also chapter 9 for injury due to chronic mechanical exposure).

Mild traumatic brain injury (MTBI) is defined as a complex pathophysiologic process induced by mechanical loading of the brain. Typically MTBI is associated with a range of clinical symptoms that are common with those observed in mild diffuse cerebral injury (see section 3.2). The symptoms that can include temporary impairment of neurological functions usually recover after a few days. Nonetheless, MTBI must be regarded as an injury of the brain that requires treatment and monitoring, particularly since repeated MTBI is believed to result in chronic degenerative brain damage [e.g. Biasca et al. 2006a, b]. Therefore it is recommended to document every MTBI. Guidelines and tools for standardized documentation of MTBI (e.g. the Sport Concussion Assessment Tool) are developed by different sports associations like the Concussions in Sports Group of IOC (Intern. Olympic Committee, FIFA (Fédéral Intern. Football Association) and IIHF (Intern. Ice Hockey Federation).

Several studies are presented that address possible injury criteria and threshold for MBTI. In a general approach, using an advanced finite element model of the head, Zhang et al. (2004) investigated mild traumatic brain injury sustained in professional football games. The injury predictors and injury levels were analysed based on resulting brain tissue responses and were correlated with the site and occurrence of MTBI. Predictions indicated that shear stress around the brainstem region could be an injury predictor for concussion. The induced shear stress may alter brain function

leading to injury. A shear stress of 7.8 kPa was proposed as the tolerance level for a 50% probability of sustaining a MTBI. If the head was exposed to a combined translational and rotational acceleration (impact duration of between 10 to 30 ms) the suggested tolerance for reversible brain injury level was less than 85g for translational acceleration. For the rotational acceleration, it was less than 6000 rad/s^2. The proposed HIC15 value was 240. Furthermore the model indicated that intracranial pressure can serve as a global response indicator for head injury. It was found that intracranial pressure was more influenced by translational acceleration while shear stress in the central part of the brain was more sensitive to rotational acceleration.

Also boxing is an obvious source for concussion in sports, although facial injuries are the most common injuries in boxing (particularly eye injuries). Several studies estimated the loading transferred to the head by punches. Peak punch forces are reported to range from 1666 N to 6860 N whereas the figures vary strongly depending on the body weight of the boxer [Walilko 2005]. For a heavyweight boxer, Atha et al. (1985) performed experiments using a ballistic pendulum. Targeting a 7 kg cylindrical metal mass, the boxer's fist reached impact velocities up to 8.9 m/s with a resulting peak impact force of 4096 N. The peak acceleration of the pendulum was 53 g. Smith et al. (2000) determined peak loads of 4800 N for elite, 3722 N for intermediate and 2381 N for novice English boxers. Using an instrumented head form, Smith et al. (1988) measured accelerations for different types of punches to reach up to 43.6 g for translational acceleration and 675.9 rad/s^2 for rotational acceleration. Assuming a tolerance limit of 200 g for translational acceleration and 4500 rad/s^2 for rotational acceleration [Ommaya 1984, see also section 3.3] they concluded that the acceleration was below the injury threshold and consequently, they suggested that repeated sub-concussive blows were the cause for MTBI.

Walilko et al. (2005) conducted experiments in which Olympic boxers of different weight classes delivered punches to the face of an instrumented Hybrid III dummy. It emerged that the average punch force was 3427 ± 811 N, the hand velocity reached 9.14 ± 2.06 m/s and the effective punch mass 2.9 ± 2 kg. The punch force was higher for the heavier weight class due to a higher effective mass of the punch. The peak translational acceleration was 58 ±13 g, the rotational acceleration was 6343 ± 1789 rad/s^2 and the neck shear force was 994 ± 318 N. The mean HIC determined from all punches was 71. Since this value is well below the proposed NFL (US National Football League) concussion threshold of 250 (Pellman et al. 2003), it was concluded that the risk of traumatic brain injury from straight

blows inducing translational acceleration is low (less than 2%). However, the high rotational accelerations (exceeding the limit of 4500 rad/s^2) suggested an injury risk due to rotation.

For more severe brain injury like diffuse axonal injury in the white matter (DAI, cf. section 3.2), Margulies and Thibault (1992) determined the threshold for head rotational acceleration to be 9000 rad/s^2. This values is somewhat lower that those proposed by Ommaya et al. (2002) who suggested 12500 rad/s^2 for mild DAI, 15500 rad/s2 for moderate DAI and 1800 rad/s^2 for severe DAI. Comparing the load experienced by human volunteers in a boxing match to different injury thresholds, Smith and Meany (2002) concluded that boxing is unlikely to result in DAI.

Generally it can be noted that in the context of sports head injuries, rotational acceleration is suspected to play a major role with respect to the injury mechanisms for diffuse brain injury and therefore receives considerable attention. Also the consequences of heading in soccer or football are controversially discussed from this point of view [e.g. Kirkendall et al. 2001]. It is argued that the translational component of the acceleration in heading is less injurious and can more easily be resisted (e.g. by adequate neck muscle strength). Rotational acceleration in contrast is associated with a higher injury risk and should be prevented, for instance, by good technique. To reduce the risk of concussion from heading, several measures are propagated by the various sports associations ranging from appropriate exercise to prevent rotation to the use of smaller and lighter balls for juvenile players or even a ban of heading for young players.

From a biomechanical perspective, however, it seems unlikely that in sports activities either isolated linear or isolated angular acceleration is sustained. Yang (2007) for instance analysed data of the NFL and found an almost linear correlation between translational and rotational acceleration. Thus both components have to be considered when investigating the injury mechanism for head injuries.

3.6 Head injury prevention

To protect the head against injury a variety of approaches are proposed, all of which, in principle, aim at padding, load distribution and preventing head contact to an object [e.g. Newman 2002].

As described above, impact force and acceleration (both translational and rotational) are relevant physical parameters determining head loading. Most head protection devices primarily aim at reducing translational

accelerations through reduction of the forces acting upon the head during an impact; however, this often leads to a reduction of the rotational acceleration as well, since the latter are generated by impact forces acting off the centre of gravity of the head.

One possibility to protect the head is through damping of the forces that could injure the head using deformable padding materials. The stiffness of these materials, together with the available deformation distance, defines the peak force expected to act on the head, while the capability of the padding to absorb energy is the deciding factor for the duration of the acceleration pulse. Such energy absorption can be achieved by deformation or by destruction of material, like in the padding of helmets. If the moving head strikes an object that deforms and thereby allows a longer deceleration distance for the head, the forces generated will be lower. Consequently, the acceleration will be reduced and also the respective injury criteria, for example the HIC, will yield better results. The extent of energy absorption depends strongly on the material properties, the thickness and the shape of the padding (see also section 3.6.1). Thus to effectively protect the head by padding, a compromise between the following requirements has to be found: allowable padding thickness, maximum padding area, uniformity of the crushing strength of the padding material, and weight.

The first two requirements address the energy absorption capabilities, while the force developed during the impact is controlled by the third. Lastly, the duration of the impact acceleration pulse is also controlled by the elasticity of the material.

For practical reasons, the range of most of these parameters is limited. Basically the same principles as for e.g. padding of the car interior apply to helmet design, but additional requirements, for instance concerned with comfort or ventilation, have to be taken into account. Additionally, controlling the weight of helmets through selection of weight-effective padding materials is paramount. Consequently, various guidelines and regulations are available defining the requirements for different types of helmets, like motorcycle or sports helmets.

In addition to energy absorbing elements as mentioned, all modern vehicles are equipped with restraint systems such as seat belts and airbag systems for head injury protection.

The seat belt (i.e. a three-point belt rather than a lap belt) aims at preventing head contact by restraining an occupant. In frontal impact, the belt effectively reduces the risk of the head impacting parts of the vehicle interior like for example the steering assembly, the A-pillar or the dashboard. Also airbags contribute to the reduction of head injury, particularly of severe brain injury [Melvin and Mertz 2002]. By distributing

the restraint force over large body areas, including the head, high force concentrations are reduced. The head benefits from smaller accelerations. Furthermore, airbags allow a better control of the deceleration of the occupant within the occupant compartment and also reduce relative motions between adjacent body parts. However, in special impact scenarios, airbags are associated with additional loading to the occupant. See chapter 5 for comments on such airbag inflation induced injuries.

3.6.1 Head injury prevention in pedestrians

Head injuries are the prevalent cause of death in car-pedestrian collisions. In order to prevent or at least mitigate the consequences of a head impact onto a car front, the deformation characteristics of the bonnet and fenders must be adapted. With the advent of homologation testing requirements in Europe, implementing compliant bonnet and fender designs has become an important task within the car industry.

Critical points are the stiff bonnet frame as well as the inner reinforcement structures. Especially, the structure of the underside reinforcements of the bonnet affects the severity of pedestrian head impact. In case the pedestrian head hits any of these stiff structures, a high head acceleration is to be expected. Lighter reinforcement structures were for instance shown to be beneficial [Kessler et al. 1988]. Recently a new aluminium bonnet without the conventional frame and reinforcement structure was presented [Mazda 2003]. This bonnet makes use of a cone-type construction to absorb and cushion impacts in the event of a pedestrian-vehicle accident (Figure 3.11). The properties of the bonnet material considerably influence the deformation behaviour, too.

Fig. 3.11 Design of a energy-absorbing bonnet [Mazda 2003].
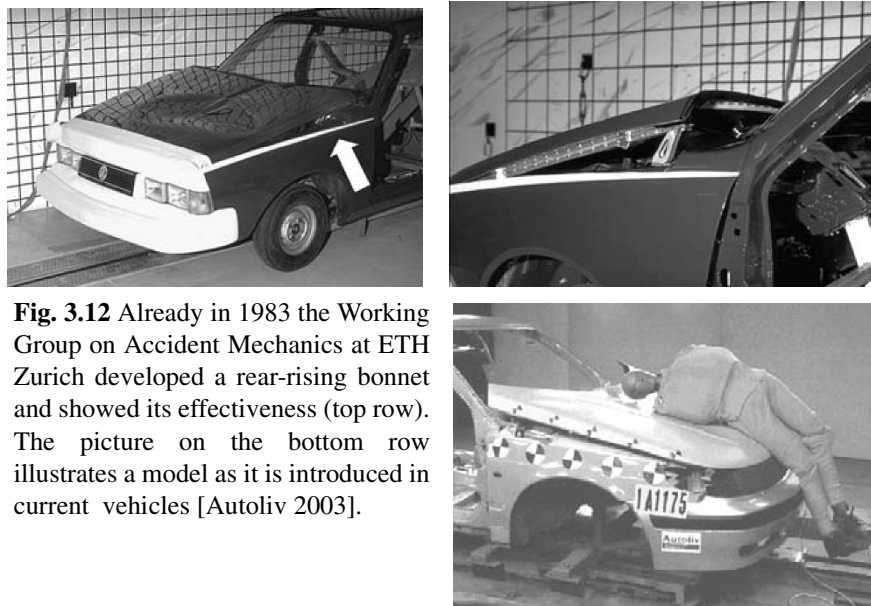
Fig. 3.12 Already in 1983 the Working Group on Accident Mechanics at ETH Zurich developed a rear-rising bonnet and showed its effectiveness (top row). The picture on the bottom row illustrates a model as it is introduced in current vehicles [Autoliv 2003].

Generally, the under-bonnet clearance is the measure that determines the maximum deformation possible. Besides the re-design of the bonnet itself, rear-rising bonnet systems (also called "pop-up" bonnets) were developed (Figure 3.12). By lifting the rear portion of the bonnet additional space underneath the bonnet is gained which can be used for deformation and energy absorption. Depending on the sensor and trigger unit used, such systems can either be activated if a pedestrian impact is likely or if it actually occurs.

Over-the-bonnet airbags and/or cowl airbags represent another injury countermeasure and may be used in addition to rear-rising bonnet systems. The over-the-bonnet airbag deploys above the bumper to cover the front of the vehicle and most of the bonnet. It is triggered by a pre-impact sensor located for example in the front grille area that detects the presence of a pedestrian in the vehicle's path, and determines whether a collision is unavoidable. The cowl airbags deploy from the base of wind shield and are triggered by an impact sensor at the front of the vehicle. These airbags together cover the cowl base of the windshield from A-pillar to A-pillar (Figure 3.13).

Finally, it is important to also consider the kinematics after the primary impact of the pedestrian on the vehicle with regard to secondary road impact (Figure 3.14). Otherwise, an optimised vehicle design to prevent injury from the primary impact might well increase the risk of sustaining

injuries from the secondary impact, which would ultimately prove to be counter-productive.
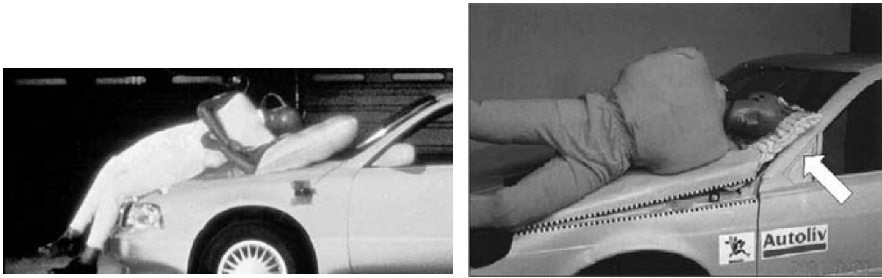
Fig. 3.13 Airbags used to prevent pedestrian injuries. Top row shows cowl airbags without (left) and with additional rear-rising bonnet (right). A vehicle equipped with over-the-bonnet and cowl airbags is shown on the right [Ford 2003, Autoliv 2003, Mazda 2003].
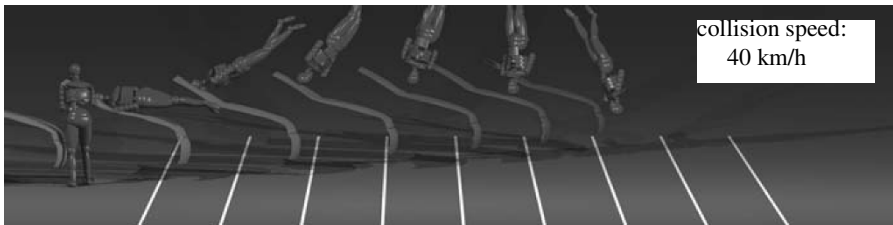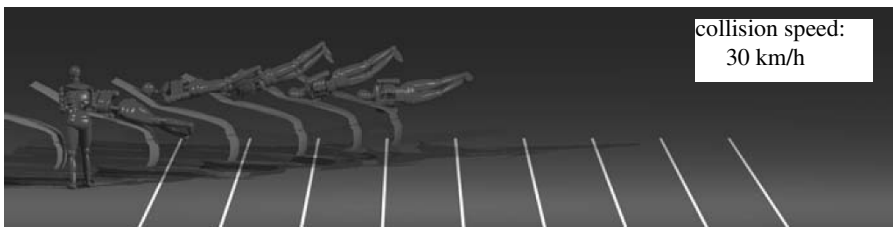
Fig. 3.14 Computer simulation of a pedestrian impact. The kinematics of the pedestrian change dramatically with increasing collision speed of the vehicle. This indicates that the secondary road impact has also to be considered [AGU 2003].

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4 Spinal Injuries

The potential for long-term impairment, including para- and quadriplegia, is always inherent in injuries to the spine and particular to the spinal cord. Of all spinal segments, the cervical spine is most frequently injured. Considering that the head and the neck form one functional entity, head loading often also implies neck loading.

With respect to traffic accidents, severe (head-contact) cervical injuries are recorded from unbelted car occupants and motorcyclists (with or without helmet). The vast majority of cervical spine injuries, however, are minor soft tissue neck injuries usually graded as AIS1 and often not exhibiting a morphological manifestation. These injuries, while not associated with overt structural injury to the cervical spine or the central nervous system, are both common and potentially debilitating. In fact, they are the most frequently occurring injuries in automotive accidents and therefore are a major concern in road traffic. In Europe, these injuries cause societal costs in the range of 5 to 10 billion Euro per year, and the number of neck injuries claimed are even increasing. Although most sufferers will make a complete recovery within a short period of time, some cases will develop prolonged medical problems placing soft tissue neck injuries among the most common causes of medical disability in car occupants [Bylund et al. 1998]. This can result in long sick leave times and granting of disability pensions. Hence, the socioeconomic significance of these injuries is tremendous. Consequently, a greater understanding of the vehicle, collision and occupant parameters that are prevalent in soft tissue injuries is needed in order to develop preventive measures.

To date, the injury mechanism underlying soft tissue neck injuries are still not yet fully understood. The medical point of view, emphasising diagnosis and treatment, is continuously discussed, but cannot be regarded as conclusive yet [e.g. Murer et al. 2002].

It should be noted that several expressions are found in the literature to describe soft tissue neck injuries. Cervical spine disorders (CSD), whiplash injuries and whiplash associated disorders (WAD) are also commonly used.

The term “whiplash” is probably used most often since it was introduced already in the last century when research first addressed this type of injury. However, one should be aware of the fact that the term “whiplash” is misleading and incorrect, because it evokes a certain injury mechanism (i.e. a forward-backward movement like during the development of the crack of a whip) although the underlying mechanism is not established (see section 4.2).

An increasing public awareness of soft tissue neck injuries and considerable attention in the media have further enforced endeavours aiming at developing “whiplash” countermeasures. Several protective devices are presented in this chapter. Other aspects of soft tissue neck injuries such as epidemiology, injury assessment, medical diagnosis and treatment, or economics related issues are not dealt with in detail here. They can be found elsewhere in the literature [e.g. Ferrari 1999, Yoganandan and Pintar 2000, McElhaney et al. 2002].

4.1 Anatomy of the spine

The human vertebral column is the principal load-bearing structure of the head and the torso (Figure 4.1). It is divided into 7 cervical, 12 thoracic, and 5 lumbar vertebrae. The vertebrae are numbered C1 to C7 for the cervical spine (with C1 being the uppermost vertebra which is connected via the occipital condyles to the skull), T1 to T12 for the thoracic vertebrae, and L1 to L5 for the lumbar vertebrae. The entire column is supported by the sacrum and the coccyx which are anatomically a part of the pelvic girdle. The size of the vertebrae increases from cranial to caudal. Adjacent vertebrae are separated by intervertebral discs. The lateral view of the entire spine shows the principal spine curves: the lordotic cervical and lumbar curves and the kyphotic thoracic curve (Figure 4.1). When viewed frontally, the normal spine is straight.

In general, each vertebra consists of a cylindrically shaped body, a vertebral (or neural) arch, the (dorsal) spinous process and transverse processes at each side (Figure 4.2). The spinous and transverse processes serve as attachment points for muscles and ligaments. Such muscles and ligaments account for stability and movements, especially of the head and neck. There are three spinal ligaments that run along the entire length of the spine: the anterior and posterior longitudinal ligaments which line the anterior and posterior aspects of the vertebral bodies, and the supraspinous ligament which joins the tips of the spinous processes (Figure 4.3). Spinal
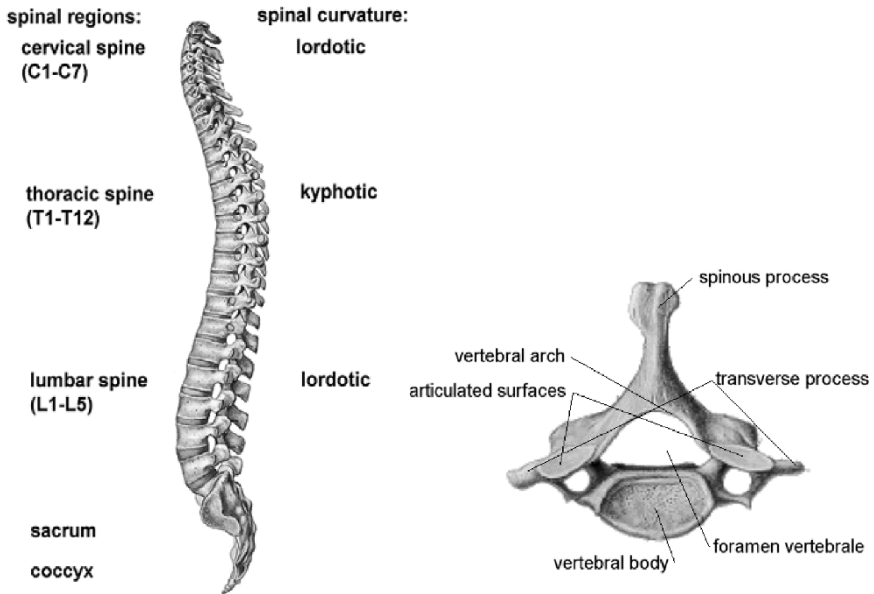
Fig. 4.1 Human spine [adapted from Sobotta 1997].

Fig. 4.2 Different regions of a vertebral body [adapted from Sobotta 1997].
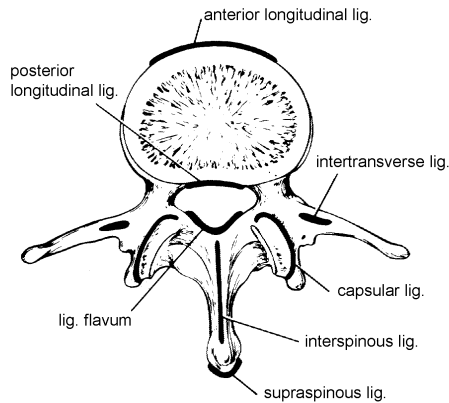
Fig. 4.3 Major spinal ligaments and their attachment points [adapted from Sobotta 1997].

ligaments are in general pre-stressed in neutral position. As for the muscles, superficial, intermediate and deep muscles are distinguished. They are symmetrical about the sagittal plane, i.e. all muscles appear in pairs. The deep muscles are closely attached to the vertebrae, while the intermediate muscles account for longer distances connecting, for example, the neck to the thorax and the skull. The superficial muscles, on the other hand, have no
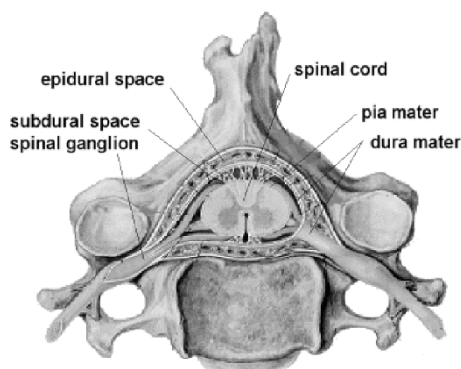
Fig. 4.4 Spinal canal and associated soft tissue [adapted from Sobotta 1997].
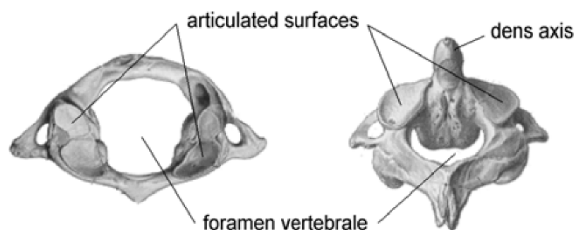
Fig. 4.5 Cervical vertebrae C1 (altas, left) and C2 (axis, right) [adapted from Sobotta 1997].

direct attachments to the spinal column.

The foramina vertebralia of all vertebrae form the spinal canal that includes the spinal cord and the associated soft tissues (Figure 4.4). The spinal cord itself is surrounded by cerebrospinal fluid (CSF). Blood vessels, especially venous vessels, are also present within the spinal canal.

In the cervical spine, it has to be noted that vertebrae C1 and C2 are anatomically different from other vertebrae. C1, also called atlas, comprises a bony ring with large articulated surfaces only. Together with the second cervical vertebra, which is characterized by a dominant process (dens axis) on its upper side, they form the atlanto-axial joint. Consequently, there is no intervertebral disc between C1 and C2 (Figure 4.5).

With respect to physiological neck motion, four basic movements are possible: flexion, extension, lateral bending and (axial) rotation (Figure 4.6) and combinations thereof. To allow this motion, different joints can be identified. Apart from the atlanto-axial joint which is responsible for head rotation, the intervertebral joints, in particular the intervertebral discs, by nature of their fibre-enforced annulus and the viscous nucleus, transmit
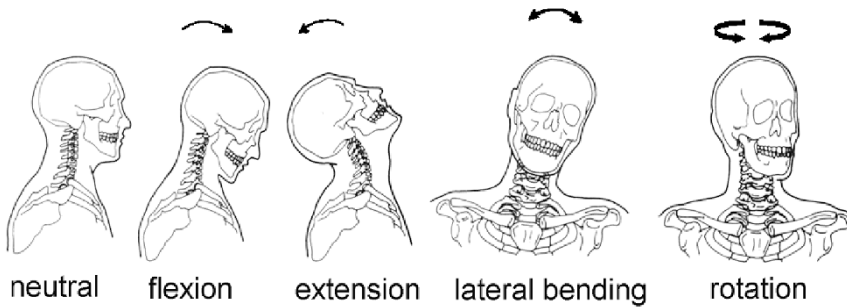
Fig. 4.6 The four basic movements of head and neck [adapted from Sances et al. 1984].

(compression and shear) forces and moments. Furthermore, the motion is guided by the two sets of facet joints (also called zygapophyseal joints) of each vertebra.

4.2 Injury mechanisms

Assessing the threat to life, the AIS (Abbreviated Injury Scale) grades several spinal injuries (Table 4.1). In general, injuries to the upper cervical spine are considered to be more serious and life threatening than those at a lower level [Viano 2001a]. In principle, injuries to the cervical spine can be classified according to the possible motion of the neck and possible mechanical loading (Figures 4.6 and 4.7).

Shear in antero-posterior direction and axial torsion may cause dislocation of the atlanto-occipital joint, while large compression might result in fracture of the atlas (C1). thereby breaking C1 into two to four sections (Jefferson's fracture, Figure 4.8). If axial compression is combined with extension of the neck, C2 fractures, commonly known as hangman's fractures, can occur. In automotive accidents this type of fracture is often related to unrestrained occupants whose forehead or face impacts, for example, the windscreen [Viano 2001a].

In automotive crashes, loading of the neck is generally due to head contact forces and combined axial or shear load with bending. Because of the anatomical curvature of the cervical spine, bending in whichever direction is almost always present. While pure compression may result in fractures as described above, non-contact head acceleration and airbag

Table 4.1 Examples of spinal injuries according to AIS scale [AAAM 2005].

AIS code	description
1	skin, muscle: abrasion, contusion (hematoma), minor laceration
2	vertebral artery: minor laceration cervical/thoracic spine: dislocation without fracture thoracic/lumbar spine: disc herniation
3	vertebral artery: major laceration cervical/thoracic spine: multiple nerve root laceration
4	cervical/thoracic spine: spinal cord contusion incomplete
5	cervical/thoracic spine: spinal cord laceration without fracture
6	decapitation cervical spine: spinal cord laceration at C3 or higher with fracture

deployment are suspected to be the cause of pure tension injuries of the upper cervical spine [McElhaney et al. 2002]. Lesions resulting from tensile loading include dislocation of the occipital condyles, ligamentous injury and fractures, e.g. dens fracture [Nightingale et al. 1998].

However, as head contact is frequently observed, the following neck injury modes are considered predominant: compression-flexion, compression-extension, tension-flexion, tension-extension and lateral bending.
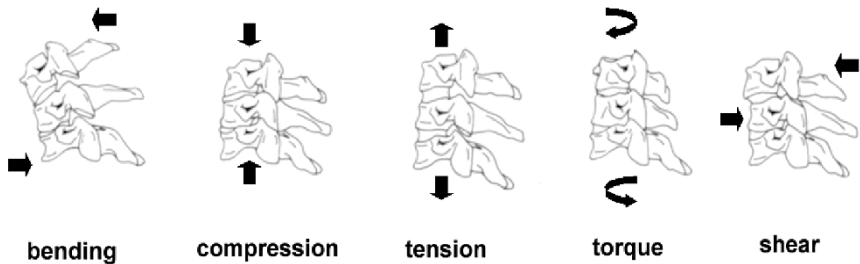
Fig. 4.7 Possible loading of the neck includes compression of the neck, tension (force stretching the neck), shear (force perpendicular to the neck column), flexion moment (forward bending of the neck), extension moment (rearward bending of the neck) and axial torsion [adapted from McElhaney et al. 2002].
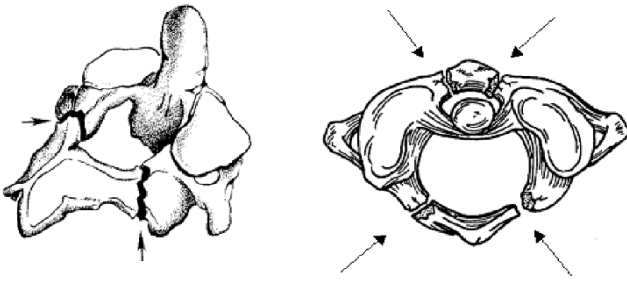
Fig. 4.8 Hangman's fracture (left) and Jefferson's fracture (right) [adapted from Vetter 2000].

Wedge fractures of the anterior vertebral bodies result from a combination of flexion bending moment and axial compression force of the neck. Head rotation often accompanies this loading scenario, but is not essential [McElhaney et al. 2002]. With increasing load, burst fractures and fracture dislocation of the facets can occur (Figure 4.9). The latter two conditions are unstable and potentially disrupt or injure the spinal cord. Hereby the extent of the injury depends on the penetration of the vertebral body or its fragments into the spinal canal [Viano 2001a].

Compression-extension loading produces fractures of the posterior structures of the neck in both the upper and lower regions [e.g. Pintar et al. 1995, Nightingale et al. 1997]. As indicated in Figure 4.10, frontal impact to the head with the neck in extension is likely to cause compression-extension loading.

Frontal impact where the torso is restrained and the neck is meant to stop head movement can result in flexion of the cervical spine while being
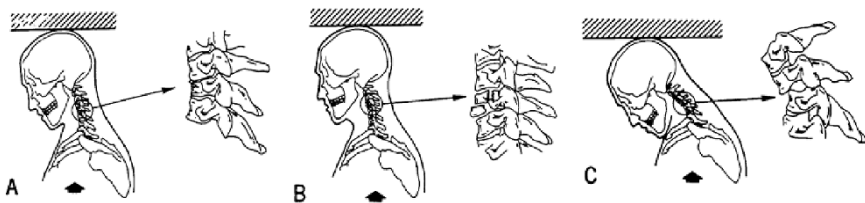
Fig. 4.9 Compression-flexion injury mechanism can result in wedge fractures (A), burst fractures (B), or bilateral facet dislocation (C). Although the figure illustrates head rotation as part of the injury mechanism, research has shown that head rotation needs not accompany this injury [adapted from McElhaney et al. 2002].

subjected to tension. Bilateral facet dislocation was observed after such loading [McElhaney et al. 2002]. However, it should be noted that such injury may also result from compression-flexion loading, suggesting that the magnitude of the bending moment rather than the axial load seems to be the determining factor.

Tension-extension loading is the underlying mechanism for several injuries (Figure 4.11). It commonly occurs when unbelted occupants hit the windscreen or when the chin impacts the dashboard. In both cases the head
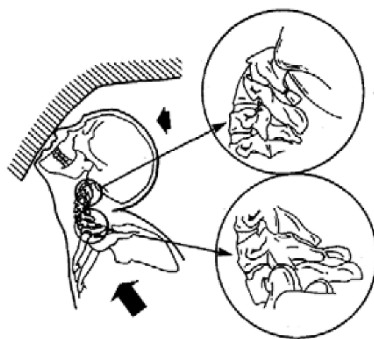
Fig. 4.10 Compression-extension mechanism. Compression of the cervical spine can be enforced by inertia forces due to the body being moving towards the head [adapted from Goldsmith and Ommaya 1984].
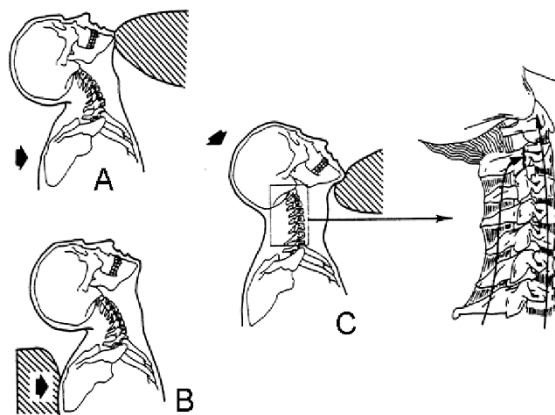
Fig. 4.11 Tension-extension loading primarily occurs through (A) fixation of the head with continued forward displacement of the body, (B) inertial loading of the neck following an abrupt forward acceleration of the torso, and (C) forceful loading below the chin directed posterosuperiorly [adapted from McElhaney et al. 2002].
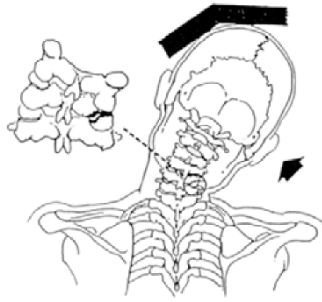
Fig. 4.12 Lateral bending and compression resulting in fracture on the compressed side [adapted from Vetter 2000].

rotates rearward and tensile force and an extension moment are applied on the neck. In case of hitting the windscreen with the head, also hangman's fractures of C2 can occur. Furthermore, soft tissue neck injuries are hypothesised to result from tension-extension loading. A more detailed discussion of this type of injury is presented later in this section.

Injuries due to lateral bending are, for instance, observed after automotive side impact. Axial loads (e.g. compression) or shear loads are often associated with lateral bending (Figure 4.12). Lateral wedge fractures of the vertebral body and fractures to the posterior structures on one side of the vertebral column can result. Additionally, lateral bending in combination with torsional loads may occur. Such cases possibly lead to unilateral facet dislocations or unilateral locked facets [Moffat et al. 1978]. However, pure torsional loads on the neck are rarely encountered in automotive accidents [Viano 2001a].

Soft tissue neck injuries are by far the most frequent injuries of the spine that are sustained in automotive accidents. Such injuries are reported from low-speed rear-end collisions, as well as from frontal and frontal-oblique collision involving belted occupants, with and without head contact. The symptoms are diverse ranging from neck pain, headache, numbness, dizziness to visual disorders and neurological deficiencies [e.g. Ferrari 1999]. In many cases no lesions are evident even if advanced diagnostic measures are applied, and therefore such injuries are most often classified as minor (AIS1). Such a classification using the AIS scale is rather broad. To allow for a more detailed assessment of soft tissue neck injuries, the Quebec Task Force [Spitzer et al. 1995] established another injury scale, categorising the symptoms and signs into four grades according to the clinical presentation (Table 4.2).

Regarding the injury mechanism, hypotheses have to rely on experiments

Table 4.2 Classification of WAD as proposed by the Quebec Task Force [adapted from Spitzer et al. 1995].

Grade	Clinical presentation
0	No complaint about the neck No physical sign(s)
1	Neck complaint of pain, stiffness or tenderness only No physical sign(s)
2	Neck complaint and musculoskeletal sign(s)
3	Neck complaint and neurologic sign(s)
4	Neck complaint and fracture or dislocation

(see section 4.3) or on symptomatic clinical observations. Due to the complex anatomy of the cervical spine various vulnerable structures are assembled in a quite small area. Hence, different tissues were proposed to be the cause for neck injuries. Many studies have suggested that ligaments and muscles are injured. The zygapophyseal joint is also often suspected to be hurt. Furthermore, nerve tissue injuries, in particular in the vicinity of the spinal ganglia, are reported. Other hypotheses can be found identifying other tissues (e.g. vertebral arteries, intervertebral discs), but these suggestions are discussed very controversially [cf. e.g. Ferrari 1999].

Analysis of the motion of the neck during rear-end or frontal collisions (both may to result in soft tissue neck injuries) reveals complex kinematic sequences including various mechanical loading conditions. The resulting movement of a car occupant during a rear-end collision (i.e. the vehicle is struck from behind and thus is accelerated forward), can for example be divided into three different phases (Figure 4.13). In the retraction phase, an occupant who is sitting upright in the driver's position is pushed forward by the seat back. The contact and force transmission occurs in the shoulder area. Due to its inertia, the head, which is not in contact with any parts of the car, has a tendency to keep its state of motion. Relative to the occupant, the head lags behind the torso. It moves back without any rotation about the lateral axis, i.e. it retracts. Hence, the upper cervical spine is forced into flexion mode and the lower cervical spine into extension. This deformation of the neck, called S-shape formation, is regarded as a crucial formation within the injury mechanism. The presence of the S-shape is well established today and was observed in various experiments using dummies
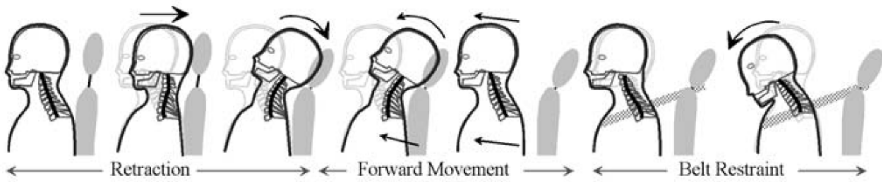
Fig. 4.13 Different phases of a rear-end collision [adapted from Muser et al. 2000].

equipped with specially designed necks as well as volunteers and cadavers [cf. e.g. Ono et al. 1997, 1998, Eichberger et al. 1998, Grauer et al. 1997, Svensson et al. 1993, Wheeler et al. 1998, Yoganandan and Pintar 2000]. Following the S-shape, the head starts rotating backwards, which in turn leads to the extension of the entire cervical spine. Eventually, the retraction phase is concluded by maximum extension. Extension depends on impact severity, the presence of a head restraint and the physiological limitations of the occupant.

The second phase is a forward movement which is characterised by a change in the direction of the movement, i.e. head, neck and torso move now forward. This phase is strongly influenced by the seat design, in particular by the elasticity of the seat and the corresponding rebound effect [Muser et al. 2000]. When the occupant returns to a position which, in sagittal direction, equals the initial position prior to the impact, the forward motion phase is completed and the belt restraint phase follows.

The belt system restrains the occupant such that the thoracic spine is stopped while the head continues to move forward. Consequently, an inverse S-shape of the neck can be observed. However, due to damping of the restraining forces by the thoracic cage, this effect is less pronounced than the S-shape experienced first. Additionally, since the (ventral) belt contact point is lower than the (dorsal) shoulder contact point, more vertebrae participate in this second S-shape formation, thus allowing for smaller loads on the individual vertebrae [Muser et al. 2000]. Finally, an overall flexion of the cervical spine concludes the kinematic phases specifically related to rear-end collision.

For frontal collisions (without head contact) similar kinematics including the inverse S-shape apply.

Considering the complex occupant kinematics, it not surprising that different mechanisms are discussed to cause soft tissue neck injuries [e.g. Walz and Muser 1995]. A shearing movement of the vertebrae has, for instance, been related to lesions of the facets of the intervertebral joints

[Yang et al. 1997]. Hyperextension of the neck, i.e. an exceedance of maximum neck moments and head excursion angles [Mertz and Patrick 1971] is also considered to be important possibilities. This mechanism is erroneoulsy still mentioned in many physician's reports although with current head restraints such a mechanism has become rare. Additionally, a hypothesis proposed by Aldman (1986) takes into consideration a pressure gradient that develops in the venous and cerebrospinal fluid of the spinal canal causing cellular injuries [Svensson et al. 1993, Schmitt et al. 2001].

Summarising, many causes for soft tissue neck injuries are hypothesised, but the underlying mechanisms are difficult to elucidate and are therefore not fully understood, let alone scientifically proven. In practice, the S-shape deformation is regarded to play a crucial role when discussing possible injury mechanisms. Consequently, injury criteria based on the S-shape deformation have been established (see section 4.4) and new dummies have been developed that are capable of a biofidelic reproduction the relative motion between the head and the torso (see section 2.5.1). When addressing the causality of "an accident" and soft tissue neck injuries reported by a patient, a physician must be briefed about the actual technical and biomechanical collision circumstances in order to prevent a purely "medical" causality assessment based on misunderstandings and incorrect technical and biomechanical assumptions. Since cervical spine pain plays a major role in insurance and court cases it is of importance that the expert opinion is performed on an interdisciplinary basis.

Injuries to the thoracolumbar spine sustained in automotive crashes are rare and play a minor role compared to cervical spine injuries. However, back pain is commonly reported after collisions and severe injuries to the spinal cord may, of course, also occur. King (2002) identifies seven different types of thoracolumbar spine injuries: anterior wedge fractures of the vertebral bodies, burst fractures of vertebral bodies, dislocations and fracture-dislocations, rotational injuries, Chance fractures, hyperextension injuries and soft tissue injuries. With respect to automotive accidents, anterior wedge fractures result when combined flexion and axial compression loading arise. This may happen, for example, in severe frontal impacts when the shoulder harness imposes a large load across the torso, causing the curved thoracic spine to straighten out. Experiments conducted with cadavers and volunteers that were restrained by a three-point belt showed that a compressive force is generated in the thoracolumbar spine which might cause wedge fractures [Begeman et al. 1973]. Principally these injuries occur at all levels of the thoracolumbar spine, but are most likely between T10 and L2 [King 2002]. Anterior wedge fractures are also observed in aircraft accidents, particularly when ejection of the pilot is

involved. Historically, the analysis of the pilot ejection problem was the primary motivation to study injuries of the thoracolumbar spine.

Chance fractures (named after G. O. Chance who first described this type of fracture in 1948) are due to improper wearing of the lap belt in case of a frontal collision. If the angle of the lap belt relative to the horizontal plane is too flat, the belt may slide over the iliac crest, thereby compressing the abdominal organs (see also section 6.2). This also causes the lumbar spine to flex which can result in separation of the posterior elements of the spine, for instance, by ruptures of the supra- and interspinous ligaments. Furthermore, the spinal cord is stretched and might be injured.

Injuries of the soft tissue of the thoracolumbar spine are also often reported after automotive accidents. The soft tissues involved are the intervertebral discs, the various ligaments, the facet joints, the muscles and tendons attached to the vertebral column. A usual complaint of this type of injury is low back pain. Incidents provoking this complaint are manifold, ranging from minor rear-end collisions to severe frontal impacts. In some cases the back pain is associated with disc rupture or disc bulge. However, a causal relationship between an impact and a rupture does usually not exist [King 2002]. Disc ruptures are generally the result of a slow degenerative process.

4.3 Biomechanical response and tolerances

The mechanical performance of the human spine was subject to numerous volunteer, cadaver, animal and dummy tests. Experiments were conducted statically and dynamically (both with and without head impact) utilising different test set-ups (see also section 2.5). Further, the use of so-called functional units is common in spine testing. Here a functional unit usually means a motion segment consisting of two or three vertebrae. Tissue that is not of interest in the study performed (e.g. muscles) is dissected. Analysing the head-neck kinematics, some studies also use larger units that are made of a cadaver head and neck whereas the neck is fixed at its lower end and then mounted on a (mini-) sled. However, it has to be noted that the use of functional units can influence the kinematics significantly. This has to be considered when drawing conclusions from results obtained in such experiments.

Muscle activity can often not be simulated in experiments, because they are either removed (functional units) or without tonus (cadaver experiments). Only volunteer experiments offer the possibility to measure
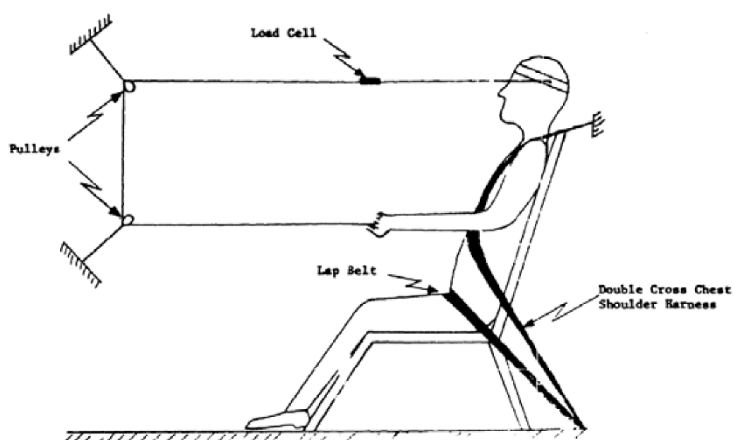
Fig. 4.14 Set-up used by Mertz and Patrick (1967) to perform static volunteer tests.

muscle activity to some extent. Tolerances to other spinal injuries, like for instance vertebral artery lesions, are difficult to assess because a physiological limit of the loading of the structure of interest cannot be defined properly.

Investigating the biomechanical response of the cervical spine, many studies still refer to and rely on tolerance levels based on volunteer and cadaver experiments that were performed in the late nineteen sixties and the early seventies, for instance by Mertz and Patrick (1967, 1971). Figure 4.14 shows a test set-up used to determine static properties of the neck.

Further, sled tests were conducted to account for the dynamic effects when loading the neck [Goldsmith and Ommaya, 1984]. Volunteer tests provided data up to the pain threshold, and cadaver tests extended the limits for serious injuries (Figure 4.15)

More recent studies investigated the relative motion of each vertebra in volunteer sled test experiments by using X-ray based techniques like cineradiography [Ono and Kaneoka 1997, 2001, Ono et al. 2006]. This way the motion pattern of each vertebra could be assessed.

For the lumbar spine, limits for frontal, rear-end and downward acceleration were proposed such that for impact durations of less than 100 ms, a 40 g threshold shall not be exceeded for well-restrained seated passengers [Viano 2001a].
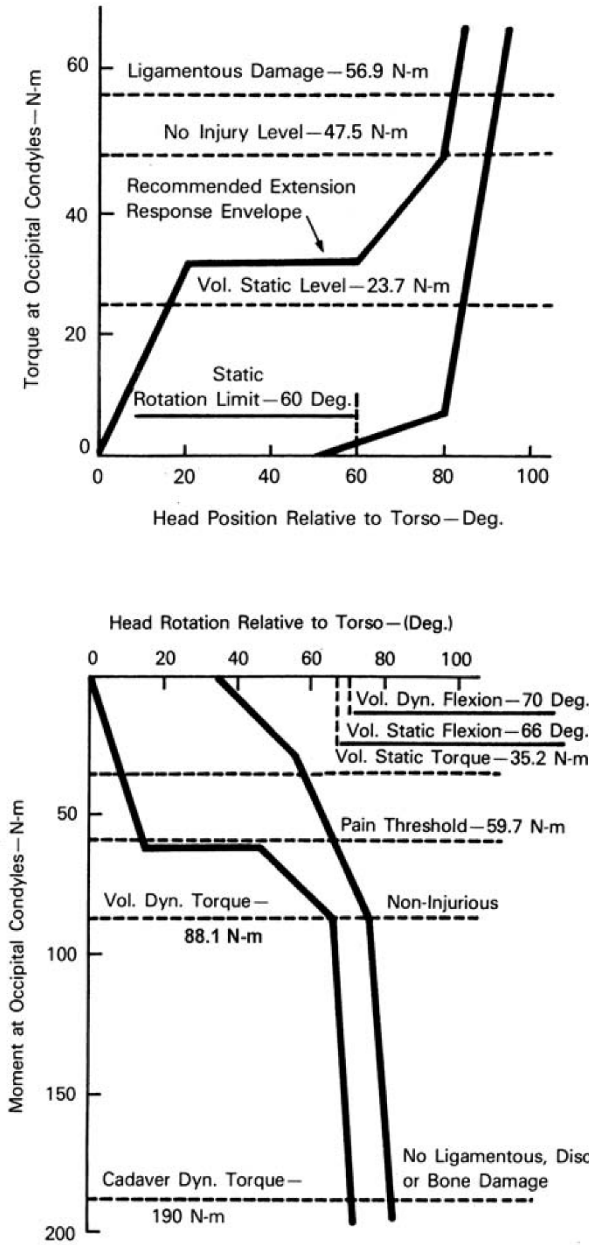
Fig. 4.15 Head-neck response envelope for extension (top) and flexion (bottom) as determined by Goldsmith and Ommaya (1984).

Table 4.3 Tolerance of the cervical spine to injury.

response measured	test objects	threshold criterion	threshold value	reference
extension	volunteers	no-injury (static)	23.7 Nm	Goldsmith & Ommaya 1984
		pain	47.3 Nm	Mertz & Patrick 1971
		no-injury	47.5 Nm	Goldsmith & Ommaya 1984
	cadavers	AIS2, ligamentous injury	56.7 Nm	Goldsmith & Ommaya 1984
flexion	volunteers	pain	59.4 Nm 59.7 Nm	Mertz & Patrick 1971 Goldsmith & Ommaya 1984
		maximum voluntary loading	87.8 Nm 88.1 Nm	Mertz & Patrick 1971 Goldsmith & Ommaya 1984
	cadavers	AIS2 (no fractures)	189 Nm 190 Nm	Mertz & Patrick 1971 Goldsmith & Ommaya 1984
compression	cadavers	bilateral facet dislocation	1.72 kN	Myers et al. 1991
		compression injuries	4.8 kN to 5.9 kN	Maiman et al. 1983
tension	volunteers	no-injury (static)	1.1 kN	Mertz & Patrick 1971
	cadavers	failure	3.1 kN	Shea et al. 1991
shear (a-p)	volunteers	no-injury	845 N	Mertz & Patrick 1971
	cadavers	irreversible damage	2 kN	Goldsmith & Ommaya 1984
	functional unit	(odontoid) fractures	1.5 kN	Doherty et al. 1993
	functional unit	ligament rupture	824 N	Fielding et al. 1974

Table 4.3 summarises tolerance values of the cervical spine derived from the various experiments reported in the literature; however, due to differences in experimental techniques and test conditions, the data shows a considerable spread. Furthermore, one has always to bear in mind that tolerance, in addition to being a function of the loading environments, is related to a variety of factors including variability of the anatomical structures (e.g. in terms of geometry or properties such as bone density) and presence of degeneration.

Hence, it is not surprising that Nightingale et al. (1997) report significant differences for male and female tolerance when testing compressive failure. Similarly, differences between the adult and the pediatric spine were found [cf. e.g. Yoganandan and Pintar 2000, Yoganandan et al. 2002]. For compression tolerance, McElhaney et al. (2002) conclude that for the cervical spine average forces of 1.68 kN and 3.03 kN result for females and males, respectively, and that tolerance values for the young human male range from 3.64 kN to 3.94 kN.

4.4 Injury criteria

In addition to the tolerance values for neck loading described in section 4.3, several neck injury criteria are defined. Besides the rather simple load limits included in current regulations, more complex criteria are proposed, particularly regarding “whiplash” injury. Even criteria used to assess the same type of collision focus on different effects, thus the assessment of an impact based on one criterion may be useful; yet it might not be sufficient. Different criteria relate to different phases and, by their definition, emphasise different aspects of the occupant motion. Hence, neck injury criteria reveal important information which can be used to describe injury risk, but sometimes also allow conclusions about associated subjects like seat design or the injury mechanism.

In general, it is important to note that injury criteria are restricted to the conditions specified in their definitions. Application to other conditions, for example to other impact directions, has to be addressed carefully. Adjustments in the test procedure and/or the evaluation and the interpretation of the results obtained might be necessary. This holds, of course, also true for the choice of the anthropomorphic test devices. As described in section 2.5.1, the designs of a dummy, and especially of a dummy neck, are very different. Therefore, when assessing neck injury by determination of neck loads and injury criteria, the influence of the dummy

is always inherent [e.g. Bortenschlager et al. 2007, Muser et al. 2000, 2002].

With respect to neck injury, the neck injury criteria NIC [Boström et al. 1996], N_{ij} [Klinich et al. 1996, Kleinberger et al. 1998], and N_{km} [Schmitt et al. 2001, 2002a] are often used. While N_{ij} was designed to detect severe neck injuries in frontal impact, the other two were developed with regard to soft tissue neck injuries in rear-end impacts. Recent work by Kullgren et al. (2003) and Muser et al. (2003) show that both NIC and N_{km} correlate well with the risk of AIS1 neck injury sustained in rear-end collisions.

A general limit of such injury criteria is the fact that they can be determined under controlled conditions, i.e. in experiments, only. Real world crashes cannot be assessed retrospectively through those criteria, because there is no possibility to measure the neck loads. With respect to soft tissue neck injuries, this poses a problem as those cases often result in legal procedures requiring an assessment by an expert witness to clarify the likeliness whether the injury claimed is causally linked to an accident. Therefore special schemes were developed to biomechanically assess this causality [Walz and Muser 2000, Schmitt et al. 2002b, 2003a]. To overcome this problem, the change of velocity (delta-v) of the vehicle under question was related to the injury risk, as the delta-v can be determined by accident reconstruction. For frontal and lateral impacts, injury thresholds ranging from a minimum delta-v of 16 km/h to 20 km/h are found in the literature [e.g. Ferrari 1999, Kornhauser et al. 1996, Watts et al. 1996, Kullgren et al. 2000]. For rear-end collisions, delta-v values of 8 km/h to 15 km/h are given [e.g. Ferrari 1999, Schuller 2001]. However, as pointed out by various researchers, it is not sufficient to take into account solely the delta-v, other vehicle specific factors such as the vehicle stiffness as well as the individual physique of an occupant have to be considered when assessing “whiplash” injury.

4.4.1 Neck injury criterion NIC

Assuming that pressure gradients caused by a sudden change of the fluid flow inside the fluid compartments of the cervical spine are related to neck injuries, the neck injury criterion NIC was developed by Boström et al. (1996). The definition of the NIC as a function of time was validated based on animal experiments. A relation to predict injury caused by pressure gradients (Equation 4.1) was found between the acceleration in the anterior-posterior direction (i.e. x-direction when using SAE J211/2) of the centre of gravity of the head relative to the first thoracic vertebra (T1) and the

velocity derived thereof.

$$NIC(t) = 0{,}2a_{rel}(t) + v_{rel}(t)^2 \tag{4.1}$$

The threshold value above which a significant risk of sustaining minor (AIS1) neck injury is assumed to be inherent was set to be 15 m^2/s^2. This value has served well in accidentological studies and is still used. However, it has emerged that reasonable values are only obtained for the retraction phase of a rear-end impact, i.e. when, in a vehicle fixed reference system, both acceleration and velocity are directed backwards. In addition, it turned out that a considerable error is introduced to the NIC(t)-curve as soon as the head is no longer parallel to T1, i.e. the head extension angle reaches values of about 20° to 30°. Thus, the NIC_{max} was introduced, which indicates that the maximum value of the NIC(t)-curve found within the time interval between the beginning of the collision and the point in time where the head, relative to the neck, reverses its direction of motion.

A modification of the NIC for low-speed frontal impact — called $NIC_{protraction}$ — has been proposed [Boström et al. 2000] and was related to long-term AIS1 neck injuries, i.e. for AIS1 injuries with symptoms for more than 6 months. As a threshold for 50% injury risk, 25 m^2/s^2 was proposed. Bohmann et al. (2000) reduced this value to 15 m^2/s^2 extending the injury assessment to short-term and long-term consequences. The following equations are used to determine $NIC_{protraction}$:

$$NIC_{generic}(t) = 0{,}2a_{rel}(t) + v_{rel}(t)\left|v_{rel}(t)\right| \tag{4.2}$$

$$NIC_{protraction}(t) = \left|Min(NIC_{generic}(t))\right| \tag{4.3}$$

4.4.2 N_{ij} neck injury criterion

This criterion was proposed by the US National Highway Traffic Safety Administration (NHTSA) [Klinch et al. 1996, Kleinberger et al. 1998] to assess severe neck injuries in frontal impacts, including those with airbag deployment and thus accounting for more severe impact conditions at higher Δv. Recently, the N_{ij} criterion was included as part of FMVSS 208.

The underlying concept for the N_{ij} can be found in a study by Prasad and Daniel (1984) who performed crash tests using piglets as child surrogates. As a result with respect to neck injuries, they suggested to combine axial forces with moments for a composite neck injury indicator. The N_{ij} criterion developed implies a linear combination of the axial forces and the flexion/extension bending moment, both normalised by critical intercept

values:

$$N_{ij} = \frac{F_z}{F_{int}} + \frac{M_y}{M_{int}} \qquad (4.4)$$

where F_z and M_y are the axial force and the sagittal bending moment, respectively. F_{int} and M_{int} indicate the according critical intercept values. These intercept values were established and validated for a three-year-old dummy. Scaling techniques were used to obtain the according intercept values for other dummy sizes and thus making the N_{ij} eligible for those dummies, too. The actual intercept values as suggested by the NTHSA are shown in Table 4.4. Hence, evaluating the criterion for all possible load cases, four different values are obtained: N_{te} for tension and extension, N_{tf} for tension and flexion as well as N_{ce} and N_{cf} giving analogue values for compression. An injury threshold value of 1.0 applies for each load case.

Adopting the N_{ij} to analyse the effect of deploying side airbags, Duma et al. (1999) replaced the sagittal bending moment by the total bending moment.

To assess AIS1 neck injury, reduced threshold values of 0.2 and 0.16 for long-term and short-term injury, respectively, were proposed [Boström et al. 2000, Bohmann et al. 2000]

However, evaluating the N_{ij} in its original form for rear-end collisions (for which it was not designed) produced difficulties in the interpretation of the results obtained [Linder et al. 2000]. Therefore a modification of the N_{ij} criterion — called the N_{km} — which is suitable for the assessment of low-speed rear-end collisions was developed.

Table 4.4 Intercept values for calculating N_{ij} as included in FMVSS 208.

Dummy	My (flexion/extension) [Nm]	Fz (compression/tension) [N]
HIII 50%	310/ 135	6160/ 6806
HIII 5%	155/ 67	3880/ 4287
HIII 5% (out of position)	155/ 61	3880/ 3880
HIII 6 year	93/ 37	2800/ 2800
HIII 3 year	68/ 27	2120/ 2120

4.4.3 Neck protection criterion N_{km}

The neck protection criterion N_{km} was proposed by Schmitt et al. (2001, 2002a). It is based on the hypothesis that such a criterion should take into account a linear combination of loads and moments. A similar approach led to the definition of the N_{ij} criterion for frontal impact [Kleinberger et al. 1998] and thus the newly proposed N_{km} can be regarded as a modification thereof.

However, with respect to possible injury mechanisms in rear-end collisions, sagittal shear forces rather than axial forces are regarded as the critical load case. A combination of shear and the sagittal bending moment accounts for a constellation often found in the cervical spine also during S-shape formation [e.g. Deng et al. 2000]. To date, the S-shape formation is mainly associated with the retraction phase, but, looking at the kinematics, an opposite S-shape, i.e. the torso lagging behind the head, could result in a similar deformation and therefore also incorporate an injury risk [Boström et al. 2000]. This opposite S-shape which can for instance be observed during the rebound phase is not assessed by the maximum NIC due to its limitations mentioned above.

Furthermore, it is assumed that shear forces could potentially be harmful to the facet joints, in particular in the upper neck region [Yang et al. 1997, Deng et al. 2000, Winkelstein et al. 2000]. Although the actual injury mechanism is unknown, the load cases of shear and extension/flexion moment seem to be relevant for neck injuries. Therefore the N_{km} does not address a single injury mechanism but takes into account a potential injury risk caused by the combination of loads and moments.

To combine shear and moment linearly seems straightforward, as for the calculation of a resulting load on a certain structure of the neck a linear combination of the existing forces and moments follows the understanding of simple mechanics. Additionally, the interpretation of the N_{km} results becomes more obvious when implementing a linear combination — a practical consideration which is important for the use of the criterion.

In the human, axial compression/tension forces are considered to influence the amount of shear [Yang et al. 1997] and are as such included. However, difficulties arise when measuring such axial forces. Performing crash tests using an ATD with standard instrumentation and an additional load cell at the upper neck position, the latter will measure the occurring axial forces. Different reasons causing an inaccuracy in such measurements can be identified:

- the centripetal force from rotation of the dummy around the pelvis is measured as an axial force.

- due to the fact that most dummies of today (except the BioRID dummy) do not represent the thoracic kyphosis, they consequently do not account for compression forces resulting from the straightening (ramping) effect.
- recent dummy designs do not allow physiological backward movement of the head during the retraction phase, i.e. when the torso is pushed forward by the seat back, the head is expected to lag behind due to its inertia. However, as the neck is connected with joints to the head and to the torso, a rotation of the head is originated which creates axial forces (although the head is intended to move relative to the torso without any rotation in the sagittal plane).
- in cases where the head in extension reaches above the head restraint and causes it to be pushed into a lower position ("hammer effect"), axial forces from this extension are also measured.

Due to this inaccuracies concerning the measurement of the axial forces, they were not explicitly included in the N_{km}. Hence, the N_{km} criterion was defined according to the following equation:

$$N_{km}(t) = \frac{F_x(t)}{F_{int}} + \frac{M_y(t)}{M_{int}} \tag{4.5}$$

where $F_x(t)$ and $M_y(t)$ are the shear force and the flexion/extension bending moment, respectively; both values should be obtained from the load cell positioned at the upper neck. F_{int} and M_{int} represent critical intercept values used for normalisation.

Distinguishing positive shear, negative shear, flexion and extension, the N_{km} criterion identifies four different load cases: N_{fa}, N_{ea}, N_{fp} and N_{ep}. The first index represents the bending moment (f: flexion, e: extension) and the second indicates the direction of the shear force (a: anterior, i.e., in positive x-direction, p: posterior, i.e., in negative x-direction). The sign convention according to SAE J211/2 was used. Consequently, positive shear forces measured at the upper neck load cell indicate that the head is moved backwards relative to the uppermost cervical vertebra.

The intercept values used to calculate the criterion are shown in Table 4.5 which exhibits the human tolerance levels for the causation of AIS 1 injuries [Goldsmith and Ommaya 1984]. These values were determined on the basis of volunteer experiments [Mertz and Patrick, 1993] and suggest tolerance levels up to which no injury is expected. The tests revealed no difference for the maximum shear tolerated in anterior and posterior direction. Sled tests as well as computational simulations have been evaluated [Schmitt et al. 2002a] to validate the proposed criterion.

For computation of the N_{km} values, the load curves measured are

Table 4.5 Intercept values for calculating N_{km}.

load case	value	reference
extension	47.5 Nm	Goldsmith and Ommaya, 1984 Mertz and Patrick, 1993
flexion	88.1 Nm	
negative and positive shear	845 N	

divided by the according intercept values, then the bending modes and the load types under investigation are identified. Finally the N_{km} values are obtained by adding the adequate shear force and moment curves, while keeping the time scale unchanged, and determining the maximum of the resulting curve. Hence, the N_{ep} for instance represents the maximum value in time when extension and negative shear occur simultaneously. If a certain combination of loads and moments is not observed within the time interval analysed, the N_{km} quadruples may be incomplete.

With regard to a critical N_{km} value, 1.0 was used taking into account that either a moment or a shear force exceeding the intercept value produces a risk of sustaining neck injuries.

To date, the N_{km} has shown its usefulness to assess low speed rear-end collisions in various tests [e.g. Muser et al. 2002, Szabo et al. 2002, Kullgren et al. 2003]. In particular, it was shown that N_{km} values allow for the characterisation of the crash phase of forward movement and as such the N_{km} gives additional information to that gained by the NIC_{max}, which accounts for the earlier phase only. As for the correlation of the N_{km} and the risk of sustaining neck injuries, Muser et al. (2003) found the N_{ea} load case to be the strongest predictor. Also Kullgren et al. (2003) report a good correlation of the N_{km} with risk of AIS1 neck injury and thus recommend to use the N_{km} (and the NIC) in rear-end test evaluation.

Furthermore, it was shown that the N_{km} values are capable to quantify different characteristics of seat design [Muser et al. 2002]. With respect to the ongoing discussion about the design principles for improved car seats, i.e. the conflict on allowing deformation (plasticity) versus elasticity [Parkin et al. 1995], the N_{km} was found to be a helpful tool since minimising both values simultaneously indicates a balanced seat design. Consequently, the criterion was proposed to be included in an ISO standard seat test procedure.

4.4.4 Intervetebral neck injury criterion (IV-NIC)

Assuming that neck pain sustained from rear-end collisions is caused by an intervertebral rotation exceeding the limit of physiological intervertebral motion, Panjabi et al. (1999) proposed the intervertebral neck injury criterion (IV-NIC). The IV-NIC is defined as the ratio of the intervertebral motion under traumatic loading θ_{trauma} and the physiological range of motion θ_{physio} (Equation 4.6). The criterion is defined for each intervertebral joint i and is calculated separately for flexion and extension.

$$IV-NIC_i = \frac{\Theta_{trauma,\,i}}{\Theta_{physio,\,i}} \tag{4.6}$$

Hence the maximum IV-NIC value identifies the time, location and bending mode of the maximal intervertebral rotation and for values greater than 1.0, it indicates that the physiological range is exceeded.
To date, the IV-NIC is neither validated nor is there a threshold value proposed. Due to the fact that in all dummy types, pin joints are used to connect the vertebrae, the intervertebral motion cannot be mimicked, and therefore the evaluation of the IV-NIC is impossible in ATD experiments. Difficulties also arise in defining the physiological range of motion, which was, in the study by Panjabi et al. (1999), solely defined on the basis of a single human cadaver specimen.

4.4.5 Neck displacement criterion (NDC)

The neck displacement criterion (NDC) has been proposed to assess the risk of soft tissue neck injury [Viano 2001b]. It addresses the S-shape of the neck by taking into account the extension moment, displacement in z (axial) direction and displacement in x (sagittal) direction. By plotting the head rotation versus the x-displacement and plotting the z-displacement versus the x-displacement, two NDC diagrams are obtained. Sled tests utilising volunteers, the BioRID and the Hybrid III dummy were performed to define tolerance corridors for the NDC diagrams. However, these corridors cannot yet be regarded as definitely set. A study presented by Kullgren et al. (2003) concluded that the NDC does not correlate well with the real life risk of sustaining “whiplash” injury. Until today no additional substantial work concerning the NDC is published and the criterion is hardly used.

4.4.6 Lower Neck Load Index (LNL)

A further neck injury criterion to assess the risk of soft tissue neck injuries, called Lower Neck Load Index (LNL) was proposed by Heitplatz et al. (2003). The LNL takes into account three force components and two of the moment components measured at the base of the neck (equation 4.7). Hence, evaluation of this criterion requires a dummy that is equipped with a lower neck load cell.

$$LNL(t) = \left| \frac{\sqrt{(My_{lower}(t))^2 + (Mx_{lower}(t))^2}}{C_{moment}} \right| + \left| \frac{\sqrt{(Fy_{lower}(t))^2 + (Fx_{lower}(t))^2}}{C_{shear}} \right| + \left| \frac{Fz_{lower}(t)}{C_{tension}} \right| \quad (4.7)$$

where $Fi(t)$ and $Mi(t)$ are the force and moment components, respectively. The denominators represent intercept values which are proposed to be $C_{moment} = 15$, $C_{shear} = 250$ and $C_{tension} = 900$ for a RID dummy [Heitplatz et al. 2003]. For other dummies intercept values are not yet proposed.

With resp t to rear-end collisions, the definition of the LNL becomes very similar to the N_{km} definition, apart from the additional term for the tension force and the fact that the data is recorded at the lower neck load cell. To date experience with the LNL is very limited. At the current stage, the LNL also comprises shortcomings such as no established biomechanical connection to a possible injury mechanism and no correlation to real world injury outcome [Bortenschlager et al. 2003].

4.4.7 Neck injury criteria in ECE and FMVSS

Current regulations specify maximum spinal loads for frontal impact (ECE R94, FMVSS 208). For low speed rear-end impact there are no homologation tests defined.

ECE R94 requires the neck extension moment not to exceed 57 Nm. Furthermore, the shear forces and the axial tension force measured should be below the values indicated in Figure 4.16.

The current FMVSS 208 includes injury criteria for the neck, consisting of individual tolerance limits for compression, tension, shear, flexion and extension moment (Table 4.6). The tolerance values are based on volunteer, cadaver and dummy tests and apply to the 50th percentile male.
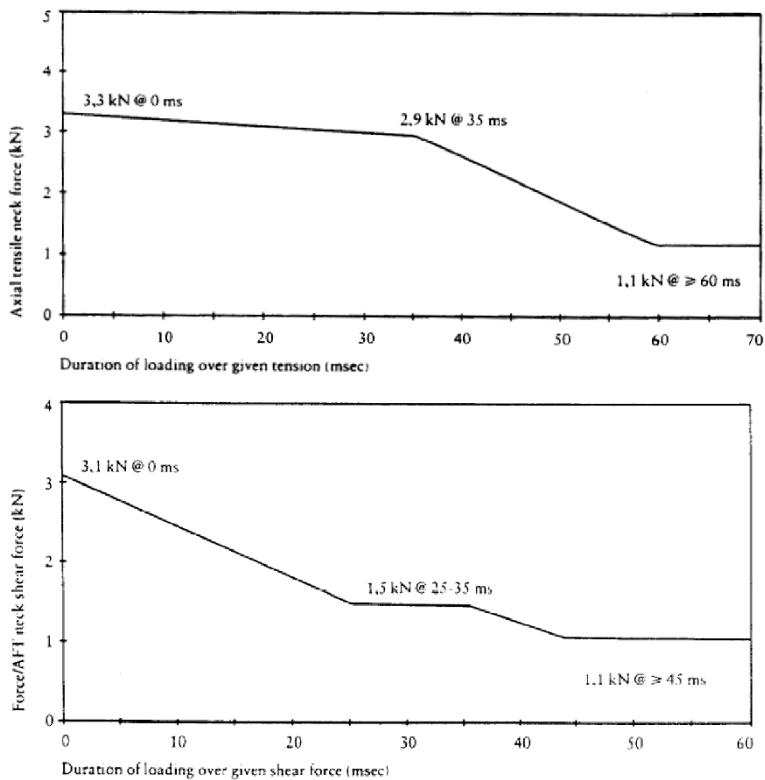
Fig. 4.16 Duration dependent limits for neck forces specified in ECE R94. Top: tension, bottom: shear.

Table 4.6 Threshold values for neck load included in FMVSS 208.

load case	threshold
flexion	190 Nm
extension	57 Nm
axial tension	3300 N
axial compression	4000 N
shear (anterior and posterior)	3100 N

4.4.8 Correlating neck injury criteria to the injury risk

As to date no AIS1 neck injury mechanism is clearly identified, it is difficult to validate the neck injury criteria proposed. Validation methods that are not related to an injury mechanism must be used to investigate the strength with which an injury criteria correlates with the real world injury risk. The information gained from such correlation analyses can then also be used for designing crash test procedures with which the risk of sustaining neck injury can be assessed.

Two studies were presented that investigated the predicitve quality of neck injury criteria with respect to AIS1 neck injury. Both studies used different methodologies and therefore serve as good examples on how the correlation of injury criteria to the real-world injury risk can be addressed.

A study by Kullgren et al. (2003) aimed at validating different proposed neck injury criteria with reconstructed real-life crashes of vehicles that were equipped with crash pulse recorders. Furthermore, the injury outcome as reported by the patients was known. A car fleet of more than 40’000 vehicles fitted with crash pulse recorders has been monitored in Sweden since 1996. All crashes with these cars, irrespective of repair cost and injury outcome have been reported. To be analysed in this study the following inclusion criteria had to be fulfilled: vehicle must be one of the three most represented car models, the crash had to be a single rear-end impact with a recorded crash pulse and the front seat occupants had to be without previous long-term AIS1 neck injury. Thus, 79 crashes with 110 front seat occupants were evaluated. In a numerical simulation study the seats of the three chosen car models were exposed to the recorded crash pulses. A model of a BioRID was used to represent the occupant. The dummy readings were correlated to the real-life injury outcome. The effectiveness to predict AIS1 neck injury was assessed for the NIC_{max}, the N_{km}, the NDC and the lower neck moment.

Concerning vehicle acceleration as injury predictor, Kullgren et al. found that below 5 g mean acceleration, the risk to sustain a long-term neck injury (i.e. the symptoms last longer than one month) appeared to be very low. At mean accelerations above 7 g the risk seems to approach 100%. No one was, however, observed to have symptoms for more than one month as long as the mean acceleration was below 3 g.

Furthermore, it was found that the NIC_{max} and the N_{km} are applicable to predict risk of AIS1 neck injury when using a BioRID as human surrogate. Together with the statistical analysis showing relatively high positive predictive values and very high negative predictive values for both NIC_{max} and N_{km}, these facts indicate that both injury criteria separately influences

injury risk. Therefore both criteria could be used to predict the neck injury risk and thus it was suggested that both NIC_{max} and N_{km} should be considered when evaluating rear-impact crash tests.

$$MIX = \sqrt{\left(\frac{NIC_{max}}{NIC_{av}}\right)^2 + \left(\frac{N_{km}}{N_{av}}\right)^2} \tag{4.8}$$

As a first attempt to combine these criteria the MIX criteria was developed (equation 4.8). Here the index "av" means the average NIC_{max} and N_{km}, respectively, of the sample analysed. The MIX was found to be useful to predict neck injury, but further studies should be conducted in this area.

Regarding the NDC no correlation to injury outcome was seen. This was partly explained by the fact, that the NDC was developed to predict AIS1 neck injury when using a Hybrid III dummy while this study used the BioRID to represent the human. Also the lower neck moment was found to be less applicable.

The second study is by Muser et al. (2003). Here the correlation of neck injury criteria and the real world injury risk was assessed on the basis of results from sled test experiments and an accident data base. The sled tests were performed with various seats from current car models. Two different anthropomorphic test devices were used: a BioRID and a RID2. After evaluating the sled tests, the results obtained were correlated with the risk to sustain whiplash injury as determined from an accident data base. The correlation was evaluated graphically by plotting linear trend lines for each dummy type and for each parameter analysed.

The data base used in this study was established by the GDV Institute of Vehicle Safety, Munich, and served as a basis to determine the neck injury risk in real world accidents. This data base registers collisions reported to a large German insurance company and was evaluated with respect to rear-end accidents that occurred in the years 2000 and 2001 (N=approx. 300'000). To qualify for inclusion in this study a case had to fulfil the following requirements:

- collision type: rear-end collision
- according sled test data must be available for comparison
- minimum of 100 cases per car model

Evaluating the data base, five different car models were identified which qualified for entry in this study. Different car models were separated and the according neck injury incidence rate was compared. The neck injury criteria as measured in the sled tests were then correlated to the real-world injury risk by plotting linear trend lines for each dummy type and for each parameter analysed. With this method it was possible to predict the

protective potential of a seat system. As a result, the N_{ea} load case of the N_{km}, when applied with a BioRID dummy, showed the strongest correlation to the injury risk. Also the NIC_{max} showed in all cases a positive correlation between injury risk and values measured in the sled tests. However, due to the small number of cases currently registered in the data base, it was concluded that it is far too early to include or exclude some of the existing criteria from further consideration.

4.5 Spinal injuries in sports

Spinal injuries that result from sports accidents comply with the same principles as mentioned above. Additional, direct blows to the spine are observed.

Most vulnerable is the cervical spine; most often the underlying injury mechanism is a compression-flexion mechanism (Figure 4.9). In neutral position the cervical spine exhibits an extension due to normal lordosis. When flexing the neck (to approx. 30°), the cervical spine straightens. If a force is now applied to the vertex, the load is transmitted along the longitudinal axis of the cervical spine without much energy being dissipated by the paravertebral muscles. Hence the cervical spine is compressed between head and torso. Fracture, luxation or dislocation may result. Examples for such a mechanism include headfirst techniques in American football and contact sports as well as diving accidents.

The latter are frequently reported from diving into shallow water, often in conjunction with a head impact, but can also occur in high diving. Commonly injuries to the cervical spinal at the level of fifth and sixth vertebra are observed [Aito et al. 2005].

Fortunately, catastrophic cervical spinal cord injuries are relatively uncommon during athletic participation. Since 1945, 497 players died in American football in the US of which 16% were due to spinal cord injury (SCI) [McIntosh and McCroy, 2005]. Stinger and transient quadriplegia/ paresis, in contrast, are more frequent injuries that have a wide spectrum of clinical severity and disabilities [e.g. Vaccaro et al. 2002].

Concerning the cervical spinal cord, episodes of transient quadriplegia are reported whereas the episode is usually followed by complete recovery occurring in ten to fifteen minutes, but sometimes taking up to two days [Torg et al. 2002]. Neurapraxia is classified according to the type of neurologic deficit. The term plegia is used for episodes with complete paralysis; paresis for episodes with motor weakness and paresthesia for

episodes that involved only sensory changes without motor involvement.

In athletes with diminution of the anterposterior diameter of the spinal canal, the cord can, on forced hyperextension or hyperflexion, be compressed, causing such transient motor and sensory manifestations. The mechanics of cervical spinal cord compression were described by Penning (1962) as "pincer mechanism". Pavlov et al. (1987) devised the measurement of the spinal canal to vertebral body ratio to determine whether an athlete has a narrow spinal canal and therefore exhibits a higher risk for cord compression. The spinal canal to vertebral body ratio is described by the distance from the midpoint of the posterior aspect of the vertebral body to the nearest point on the corresponding spinolaminar line divided by the anteroposterior width of the vertebral body. Normally, regardless of gender or age, the spinal canal to vertebral body ratio is close to one. A ratio of smaller or equal 0.8 was recorded at one or more levels in patients who experienced cervical cord neurapraxia.

With respect to other segments of the spine, low-back pain is a symptom for which a life time prevalence in the general adult population of 85% - 90% is estimated [Trainor and Wiesel 2002]. Hence, also athletes suffer from low-back pain although it is not clear whether they are at higher risk [Bono 2004]. Some studies suggest that for certain athletes (like wrestlers and elite gymnasts) there might be a higher risk, but results are not yet conclusive. Various risk factors are investigated taking into account lumbar flexibility, lower-extremity function or the footwear used. So far, a history of low-back pain was found to be the greatest predictor of future occurrences of low-back pain in athletes [Bono 2004].

Participation in sports appears to be a risk factor for the development of disc degeneration with disc degeneration being influenced by the type and intensity of the sport [Sward et al. 1991]. The prevalence of spondylolysis in athletes was found not to be higher than that in the general population. However, some studies suggest that there seems to be a higher prevalence in some sports like weight lifting, diving, wrestling [Bono 2004].

Sacral stress fractures are reported almost exclusively in high-level running sports such as marathon (see also chapter 9 for stress fractures).

4.6 Prevention of soft tissue neck injury

Since causes for neck injuries sustained in sports are manifold depending, for instance, on the type of sports, the condition of the athlete and the actual situation, it is hardly possible to apply a general strategy for prevention.

Thus the following discussion is limited to soft tissue neck injuries sustained in automotive accidents.

Recent developments in vehicle seat design aim at providing seats that offer good "whiplash" protection. This proves to be a rather difficult task, given that the underlying injury mechanisms are not known. In a holistic approach to prevent "whiplash" injury, general design guidelines for seat development were established [Walz and Muser 1995, Lundell et al. 1998]. Such guidelines attempt to address all existing hypotheses concerning the injury mechanism by minimising relative motion between head and thorax and thus reducing all kinds of biomechanical loading that might cause injury. There is a certain risk in this approach that much time and effort might be spent in reducing certain neck loads that are not responsible for WAD at all, or in reducing loads that were sub-critical already before the improvement process started. The ever increasing case numbers and associated societal cost demand, however, that measures be taken despite this risk, since it might still take a long time for researchers to solve the biomechanical problems associated with soft tissue neck injury.

It is assumed that without relative acceleration between head and torso, no soft tissue neck injury will be sustained. Bearing in mind that the neck injury criterion NIC is widely used to assess the risk of soft tissue neck injuries, it is especially this relative acceleration that has to be reduced to obtain good NIC values. Taking into account hypotheses that claim the relative movement between adjacent vertebrae to be causal for "whiplash" injury, such motion must be avoided. Hence, the curvature of the spine must be kept unchanged during the impact. Additionally, the rebound phase has to be considered. To minimise the interaction with the seat belt, rebound must be reduced.

The potential of a seat to prevent soft tissue neck injury is assessed by performing experiments where accelerations, forces, and moments of torque and determining various neck injury criteria thereof are measured. To ensure a broad basis for analysis and assessment, none of the measures mentioned can account for all factors alone, but all should be reduced. Due to the uncertainty with respect to the injury mechanism, an increase of any response related to the biomechanical guidelines - even if accompanied by a clear reduction of another criterion - is to be avoided.

Today, several seat systems to prevent "whiplash" injury are available on the market. Basically all major structures of a vehicle seat such as the head restraint, the seatback including the recliner, as well as the seat base and the seat slide serve as the basis for the development of different "whiplash" protection devices. The following sections summarise the most recent advancements.

4.6.1 Head restraint geometry and padding material

Head restraints, originally introduced to prevent severe injuries to the neck related to hyperextension, may also prevent the sudden relative movement of head vs. torso (cf. S-shape). This protection potential can only be exploited if the head restraint is positioned correctly.

The influence of the head restraint geometry on the protective potential of the head restraint was investigated in several studies. A decrease in "whiplash" injury incidence was found with increasing head restraint height [Eichberger et al. 1996, Hell et al. 1998, Ferrari 1999]. Similarly it was found that the head to head restraint distance is associated with injury potential [Ferrari 1999, Hofinger et a. 1999, Wiklund and Larsson 1998]. The shorter the head to head restraint distance, the more effectively the S-shape is prevented.

Although the design of head restraints has been improving over time according to the Insurance Institute for Highway Safety [IIHS 2001], many of today's head restraints are not high enough to protect an average-sized male occupant [Chapline et al. 2000]. This may either be due to a poor head restraint geometry which does not allow a proper adjustment, or it can be due to the fact that head restraints are not adjusted correctly by the occupant. With respect to head restraint geometry, a minimum height that corresponds to the top of an average-sized male is recommended. Furthermore, the head restraint should be as close to the back of the head as possible. In addition some car manufacturers offer active or re-active head restraints which reduce the distance between head and head restraint during a rear-end impact (see 4.5.2). Add-on head restraints to minimise the head to head restraint distance are also on the market.

Not only the geometry, but also the inner structure, in particular the padding material, might be modified to prevent soft tissue neck injury. Schmitt et al. (2003a) investigated the possible effect of using energy absorbing foam as padding material in seats.

In summary, the study showed that the use of automotive visco-elastic foam clearly reduces the maximum head acceleration (Figure 4.17). A reduction of the neck injury criterion NICmax however became apparent for rear-end collisions under conditions of higher delta-v values only.

For such rear-end impacts, the numerical simulations also suggested that thicker head restraints account for reduced head and neck loading. Using the visco-elastiv foam instead of the poly-urethane foam more commonly found in car interiors, this effect was even more pronounced. Hence, the strong influence of the initial head restraint distance was confirmed.

These results corroborate a study by Szabo et al. (2002) analysing
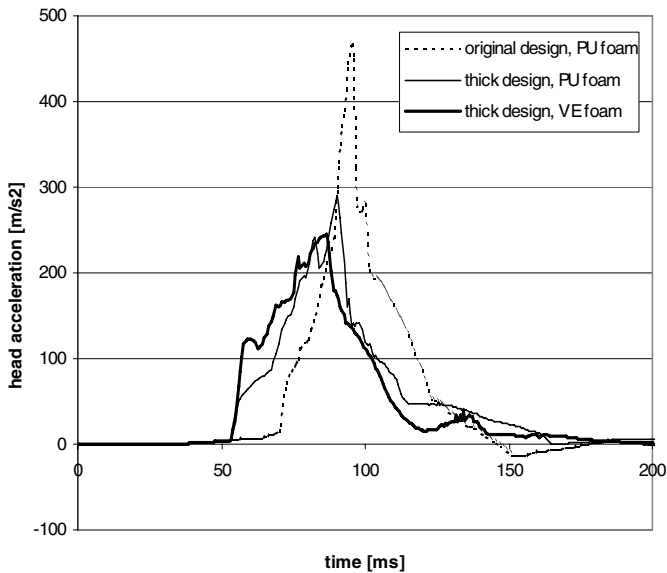
Fig. 4.17 Head acceleration as calculated with the numerical model at a delta-v of 30km/h. The peak acceleration is significantly reduced for a thick head restraint design with the automotive VE foam resulting in lower acceleration than the PU foam [Schmitt et al. 2003b].

different seats and seat back foams. Even when using visco-elastic foams it appeared that the seat geometry itself had overwhelmingly more influence on occupant kinematics and the potential of whiplash injury than the local foam properties within a given seat.

On the other hand, controlling the occupant kinematics using optimised seat back geometry along with a carefully adjusted foam stiffness distribution along the seat back and the head restraint might well prove a viable way of injury protection without any additional technical effort, and an approach used in the so-called WIL (Whiplash Injury Lessening) system [Sekizuka 1998, Sawada and Hasegawa 2005].

4.6.2 Controlling head restraint position

Theoretically, according to the 'no relative motion' paradigm presented above, WAD injuries could be avoided by simply ensuring a zero distance between head and head restraint during the collision. Several ideas have been proposed to achieve this goal. Muser et al. presented in 1994 a head

restraint equipped with capacitive sensors and electrical actuators, always keeping the head restraint at a pre-programmed distance. Mainly due to cost considerations, such a system was not introduced in the market at that time, but is available today in luxury cars.

The SAHR system [Wiklund and Larsson 1998] is a re-active head restraint system: it moves the head restraint upwards and towards closer to the occupants' head during the impact. Thus the distance between head and head restraint is reduced only when needed. Figure 4.18 illustrates the principle of the self-aligning head restraint. The rearward motion of the torso towards the seat back is used to load a plate, which in turn is connected through a lever to the head restraint. This seesaw mechanism rotates the head restraint forward, resulting in earlier head contact, and upward with respect to the occupant. Various studies demonstrated the ability of the system to prevent "whiplash" injury [e.g. Viano and Olsen 2001, Muser et al. 2002]. The SAHR system was one of the first whiplash protection systems to be included in serial production in 1997. Since then, various other, similar systems have been presented, and the design principle has found widespread use.

Various systems that move the head restraint forward during the collision without relying on the force interaction of the occupant to the seat back
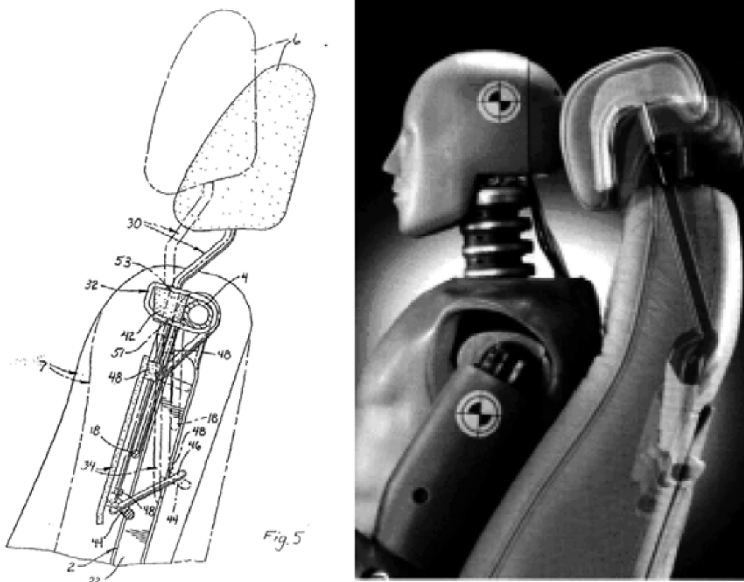
Fig. 4.18 Principle of SAHR [adapted from Viano and Olsen 2001].

have been presented as well. Some approaches relied on more or less classical airbags fitted in the head restraint, albeit inflated much slower by pressurised air. Another system (Figure 4.19) uses pre-tensioned springs as an energy source, and is also currently used in serial production. It is often argued that these active systems offer the advantage over the SAHR-types that, with the latter, an occupant having a lower body mass than e.g. the 50% male may not be able to move the head restraint far enough forward to be of any use at all.

Several aftermarket devices, essentially consisting of cushions placed between the head restraint surface and the head, also serve to reduce head to head restraint distance. Depending on the preferences of users, these devices might offer a significant protection potential; it is often argued, on the other hand, that too low a distance is not comfortable, thereby limiting the applicability of these devices to older seats with designed distances of 10 cm and more.

Fig. 4.19 Crash-active head restraint (CAK) [Keiper 2006].

4.6.3 Controlling seat back motion

Another approach is based on the insight that not the relative motion as such, but rather its 'violence' in terms of relative velocity and acceleration, generates the risk of WAD injury. It should therefore be possible to lower injury risk by e.g. damping devices in the seat back that, first, lower the acceleration at the upper torso level and, consequently, also the relative velocity between head and torso.

The WHIPS seat [Lundell et al. 1998] is equipped with a recliner that allows controlled backward movement of the seatback during rear-end impact. If a critical load is exceeded, the motion is performed in two steps: a translational rearward movement of the seatback is followed by a
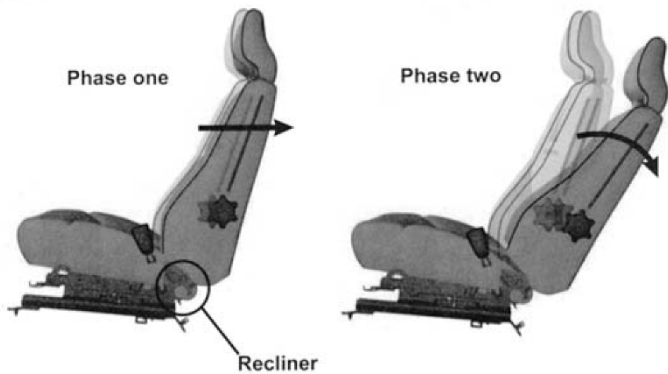
Fig. 4.20 The WHIPS seat allows first a translational rearwards motion which then is followed by a rotation of the seatback [adapted from Lundell et al. 1998].

rotational motion reclining the seatback (Figure 4.20). Advantageous neck injury criteria values were observed for the WHIPS seat tested under low-speed rear-end conditions [e.g. Hell et al. 1999, Muser et al. 2000, Langwieder et al. 2000]. The WHIPS system is also among the first systems that appeared on the market, allowing a statistical analysis of its efficiency in preventing WAD [Jakobsson 2005].
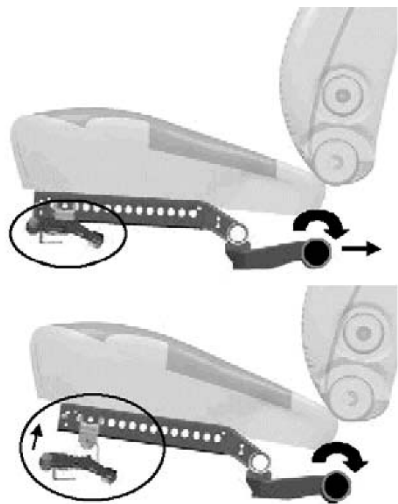
Fig. 4.21 The WipGARD systems also allows rotational and translational motion (top). To lift the seat base a rivet must be torn (bottom) [adapted from Zellmer et al. 2001].

Similar effects may also be achieved by interfering with other parts of the seat. An idea presented by Schmitt and Muser [Schmitt and Muser 2002, Schmitt et al. 2003c, 2003d] is based on the seat slide present in every car to allow the adjustment of the longitudinal seat position. Assuming that the relative acceleration between head and T1, and consequently also the NICmax value, is to be reduced to prevent whiplash injury, a device was developed which allows a translational motion of the seat relative to the car while damping this motion. This leads to a delay in the torso acceleration, thereby synchronising the loading of the head and the upper torso.

Sled tests and numerical simulations have shown that already a relatively small deformation distance of the seat slide (approximately 40 mm) is sufficient to significantly reduce injury criteria values such as NIC and Nkm. From an engineering standpoint, changes in the seat slide are much simpler to implement than changes in e.g. the recliner, since the slide is much more easily exchanged.

Therefore, also the WipGARD [Zellmer et al. 2001], an aftermarket device which can be fitted into certain Volkswagen models, is mounted between the seat slide and the floor. Like WHIPS, WipGARD enables the seat back to perform a translation followed by a rotation. WipGARD, however, allows the entire seat to move in the prescribed manner (Figure 4.21). To activate the system, WipGARD requires a critical load.

With the various above mentioned systems being introduced more and more in the market, first results from accidentology already indicate that some systems have a significant positive effect to reduce the injury risk [Jakobsson and Norin 2004, Krafft et al. 2004]

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5 Thoracic Injuries

Injury to the thorax commonly occurs in impact from the front and the side as well as in all impact directions intermediate to these two. Impact to the thorax is frequently observed due to contact, for example, with various components of the vehicle interior (like the steering assembly, the door or the dashboard) or an opponent player in sports.

Most thorax injuries caused by contact mechanisms are due to blunt impact. In automotive accidents, sharp impact to the thorax is rare, occurring only due to obstacles inside the passenger compartment or when the occupant is ejected from the vehicle. Consequently, this chapter focuses on blunt impact.

5.1 Anatomy of the thorax

The thorax consists of the rib cage and the underlying soft tissue organs. It extends from the base of the neck to the diaphragm which inferiorly bounds the thorax and separates the thoracic cavity from the abdominal cavity (Figure 5.1).

The rib cage is formed by twelve pairs of ribs which are posterior connected to the thoracic vertebrae of the vertebral column. At the anterior side of the thorax the sternum fixes the upper seven ribs. The lower ribs are either connected indirectly to the sternum or are attached to muscles and the abdominal wall (so-called floating ribs). The ribs are interconnected with each other by the internal and external intercostal muscles. As the connections of the ribs to the vertebrae, the intercostal muscles, and the sternum are flexible, the rib cage represents a quite stiff though deformable cover of the internal organs and facilitates respiration.
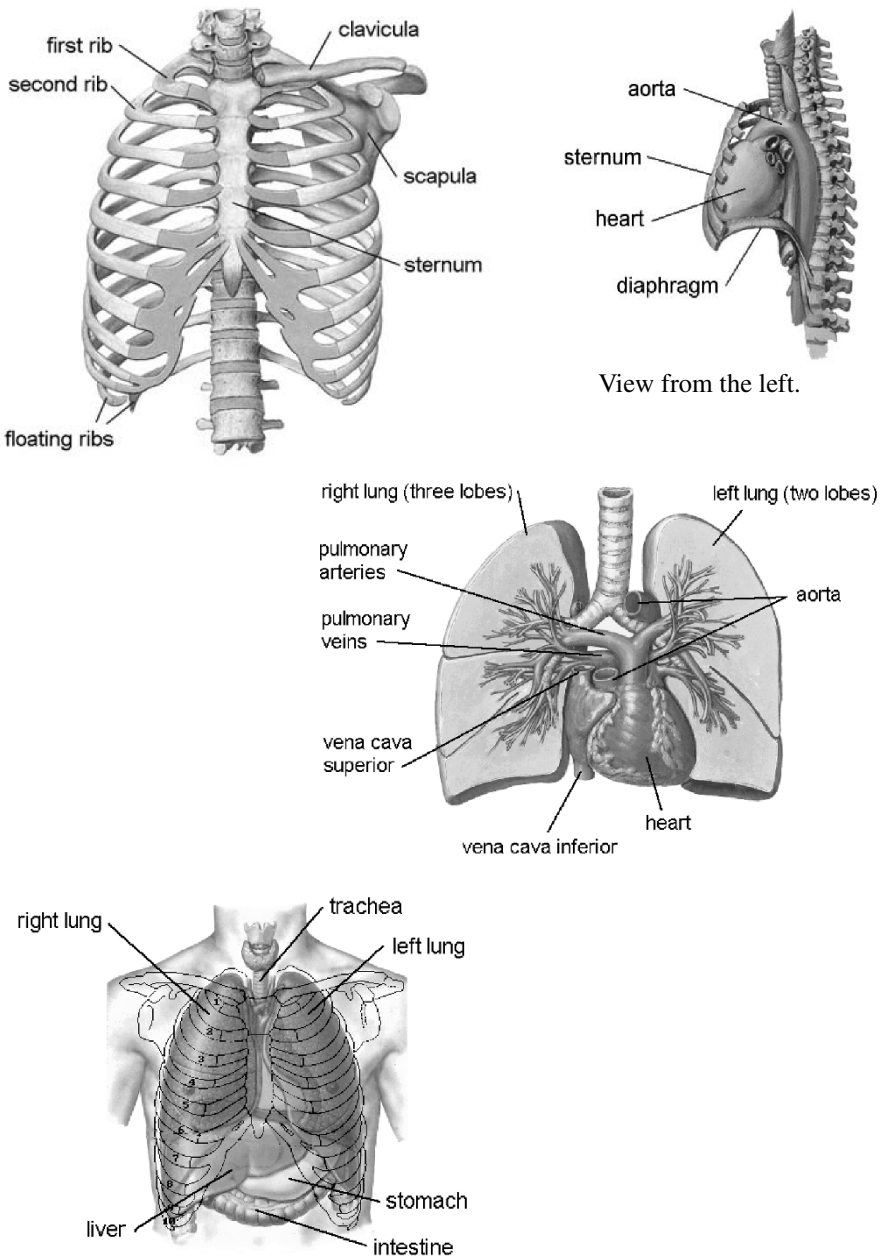
Fig. 5.1 The thoracic anatomy [adapted from Sobotta 1997, Netter 2003].

While the rib cage is very flexible in a new born, the stiffness increases with growth but still retains a certain flexibility. In the elderly the joints between the ribs and the sternum and the vertebrae, respectively, become stiffer. In addition, the ribs become more brittle due to changes of the bone properties. This increases the likeliness of rib fractures and reduces the protective potential of the rib cage.

The interior volume covered by the rib cage can be divided into three areas. The right and the left outer region contain the lungs. The centre section, called mediastinum, hosts among others the heart, the trachea and large vessels.

The left lung consists of two lobes, while the right lung consists of three lobes. Two layers of membranes surround the lung: the visceral pleura, which encloses the lung tissue, and the parietal pleura, which covers the entire inside of the rib cage (including the cranial side of the diaphragm and the vertebral bodies). The visceral and the parietal pleura are not connected to each other, but form a small cavity. This pleural cavity is an enclosed space. To keep the lung in their inflated state, a continuous underpressure is maintained in the pleural cavity. If this underpressure cannot be maintained (for example due to a perforation of the chest), the lungs will deflate and the pleural cavity is filled with air. This phenomenon is called pneumothorax (see section 5.2.2).

For respiration the diaphragm, the rib cage and the intercostal muscles function as a pump by drawing air into the lung (inspiration) and expelling air from the lung (expiration). For inspiration the thoracic volume is increased by lifting the rib cage and by lowering the diaphragm. Consequently the lung will expand and air is sucked in. To normally expel the air (expiration) the thoracic structures and the diaphragm are relaxed.

The mediastinum is located between the two lungs, the thoracic vertebrae and the sternum. Large vessels included are the aorta, vena cava, the pulmonary arteries and veins (Figure 5.1). Due to the restricted space available in the mediastinum, a compression of the anterior rib cage may easily cause injuries to internal structures.

5.2 Injury mechanisms

This description of thoracic injuries and the according injury mechanisms focuses on blunt impacts in traffic accidents. Hence only scenarios where a flat or blunt object strikes the chest without penetration are regarded. This type of impact is most often seen in automotive accidents with an occupant

contacting, for example, the steering wheel, the dashboard or components of restraint systems.

If the thorax is suddenly decelerated due to a blunt impact, three different injury mechanisms can be distinguished: compression, viscous loading and inertia loading of the internal organs. Furthermore any combination of these three basic phenomena can occur.

The resulting injuries can be categorised as skeletal injury and soft tissue injury. Most often the thoracic wall and the lung are injured together with rib fractures, fractures of the sternum, and pleura ruptures. In case of fractures of the vertebral column, injuries of the spinal cord may also occur, possibly leading to transverse lesion (In case of motorcyclists this results sometimes in quadriplegia). Fortunately such injuries are less frequently recorded, as are injuries on the aorta, the heart, the oesophagus and the diaphragm. Table 5.1 provides an overview of different injuries and their according AIS (Abbreviated Injury Scale) rating. However, it has to be noted that injury statistics based on the current vehicle fleet still contain a significant proportion of vehicles not equipped with advanced restraint systems. Furthermore, a significant proportion of car occupants is still not or not correctly using restraint systems such as the seat belt. Hence, with the increased use of seat belts and the requirement for advanced frontal airbags, the proportion of skeletal injuries that were often sustained when

Table 5.1 AIS rating for skeletal and soft tissue thoracic injuries [AAAM 2005].

AIS	Skeletal Injury	AIS	Soft tissue injury
1	one rib fracture	1	contusion of bronchus
2	2-3 rib fractures; sternum fracture	2	partial thickness bronchus tear
3	4 or more rib fractures on one side; 2-3 rib fractures with hemothorax or pneumothorax	3	lung contusion; minor heart contusion
4	flail chest; 4 or more rib fractures on each of two sides; 4 or more rib fractures with hemo- or pneumothorax	4	bilateral lung laceration; minor aortic laceration; major heart contusion
5	bilateral flail chest	5	major aortic laceration; lung laceration with tension pneumothorax
		6	aortic laceration with hemorrhage not confined to mediastinum

contacting the steering assembly is likely to be lower.

To date, the mechanisms of rib fractures and some of the lung injuries are reasonably well understood whereas some mechanisms leading to other injuries of the thoracic organs still merit further research. Figure 5.2 summarises possible soft tissue thoracic injuries.
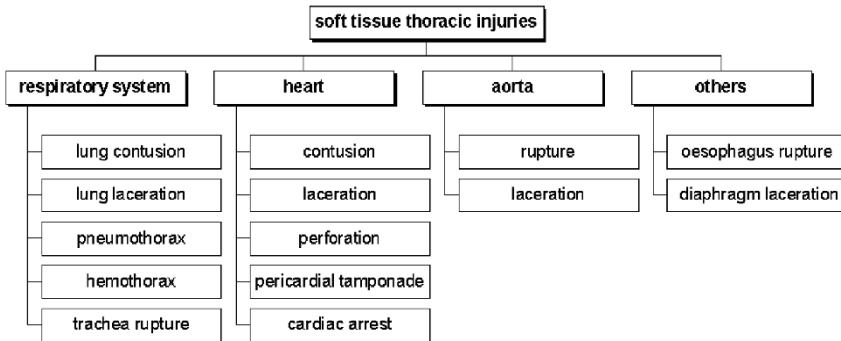
Fig. 5.2 Possible soft tissue thoracic injuries.

5.2.1 Rib fractures

According to the AIS, a single rib fracture can be graded as AIS1. If 2-3 ribs are broken, the grade increases to AIS 2. Hence this type of injury is usually not severe and most single rib fractures are in fact self healing. However, if multiple fractures occur, life threatening complications may arise. If the skin and the soft tissue overlaying the fracture remain intact, the fracture is called a closed fracture. If, on the other hand, sharp edges of broken ribs perforate the chest wall, the fracture is called an open fracture. Such open fractures are of particular concern because they can lead to a pneumothroax, lung collapse and infections. Broken ribs may also perforate the visceral or parietal pleura, causing respiratory problems.

Generally, sagittal loading of the thorax is more likely to cause single rib fracture, while lateral impact more often results in multi rib fracture. In principle, ribs can fracture at any point, but most likely they break at the point of maximum curvature or at the location where a force is applied. Hence, given the fact that the ribs are stronger curved laterally, together with fewer muscle tissue that covers and thus protects the ribs in that area, lateral fractures are more likely. The site of lateral rib fracture(s) depends on the shape of the impacting body (Figure 5.3).

In case of multi rib fracture the thorax wall may lose its overall stability.
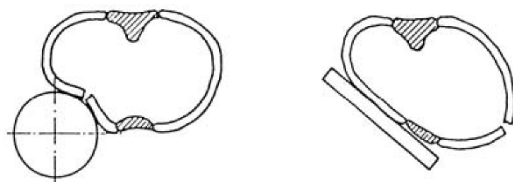
Fig. 5.3 Site of rib fracture depending on impact body [adapted from Kramer 1998].

This can result in a thorax motion that is contrary to normal: during inspiration the disrupted thorax wall is sucked in and thus reduces the volume of the lung. On expiration the thorax wall moves outwards making it difficult to expel the air out of the lung. The greater the area of thorax wall damaged, the lesser the amount of air which can be exchanged. This condition is called a flail chest which eventually results in hypoxemia.

According to cadaver studies [e.g. Stalnaker and Mohan 1974, Melvin et al. 1975] the number of rib fractures depends on the magnitude of rib deflection rather than on the rate of deflection. Due to the viscous nature of the thorax, the amount of force, however, depends on the rate at which the force is applied. Hence, force appears to be related to the number of rib fractures for a given loading rate.

The occurrence of rib fractures is strongly age dependent. While the rib cage can be compressed frontally in a young person until it contacts the spine without fracturing a rib (but compressing the organs in between), the rib cage of individuals of more than 50 years of age break at much lower loads, often for example during the cardiac rescue procedure.

5.2.2 Lung injuries

As indicated in Figure 5.2, injuries to the respiratory system mainly concern lung injuries. Due to thorax compression (both with and without rib fracture) a lung contusion can occur. This often happens in combination with a flail chest.

Unlike rib fractures, lung contusion is rate dependent [Fung and Yen 1984]. At high velocities, a compression or pressure wave is transmitted through the thorax wall to the lung tissue, causing damage to the capillary bed of the alveoli. Sometimes also central lung contusion without damage of the surrounding tissue is observed. As a serious complication, lung contusion also increases the risk of pneumonia, i.e. an inflammation of the lung tissue.

Laceration and sometimes also perforation of the lung tissue can be observed at sites of rib fractures. This may result in a pneumothorax or a hemothorax. In the first case the pleural cavity is filled with air, in the second case with blood. A combined situation where the pleural cavity contains both blood and air is called a hemo-pneumothorax.

A pneumothorax results from a perforation of the pleura, i.e. a hole is created in the pleural sac between the lung and the rib cage, caused, for instance, by broken ribs. On inspiration the intrapleural pressure is reduced and air is sucked into the pleural cavity through the leak in the lung. During expiration the laceration in the lung tissue is compressed preventing the air in the pleural cavity to be expelled. Hence, while breathing, the amount of air inside the pleural cavity increases, eventually compressing the lung.

A hemothorax also reduces the effective lung volume, but due to blood in the pleural cavity. Hereby a laceration of blood vessels (e.g. in the lung tissue) may cause blood to accumulate in the pleural cavity.

5.2.3 Injuries to other thoracic organs

From thoracic impact, the heart can be subjected to several injuries including contusion and laceration (Figure 5.2). Contusion occurs due to compression and depends on the associated velocity, while laceration may be due to high magnitude of compression over the sternum. At high rates of loading, the heart may undergo arrhythmia, fibrillation or arrest. High speed blunt impacts (15 - 20 m/s) appear to interrupt the electromechanical transduction of the heart wall. Figure 5.4 illustrates a thoracic impact, with the heart under compression between the sternum and the vertebral column.

Furthermore, major thoracic blood vessels like the aorta may be injured. Rupture and laceration are the most likely mechanisms resulting from blunt trauma to the thorax. Cavanaugh (2002) reports that arterial injuries account for 6% to 8% of AIS >2 only, but represent 27% to 30% of the estimated harm. It is also remarkable that 80% to 85% of the victims
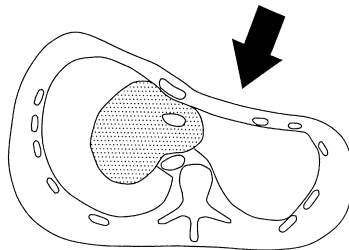
Fig. 5.4 Compression of the heart [adapted from Kramer 1998].
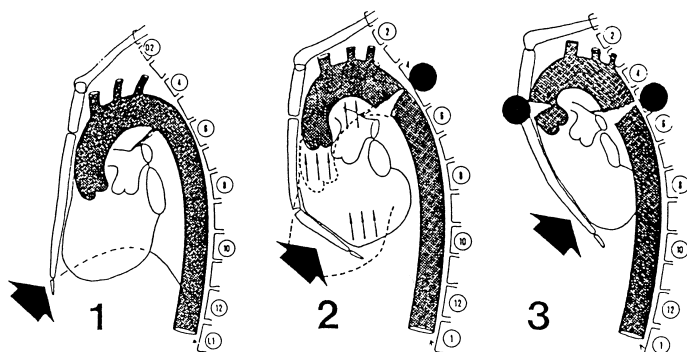
Fig. 5.5 Compression of the heart and possible sites of aortic rupture [adapted from Viano 1990].
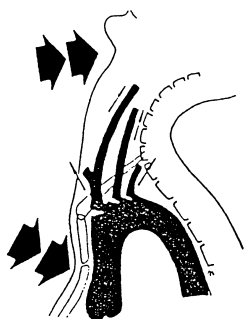
Fig. 5.6 Thorax compression in combination with hyperextension of the neck can result in laceration of the aorta [adapted from Viano 1990].

sustaining an aortic trauma in an automotive accident die at the scene of the accident [Smith and Chang 1986]. Mechanisms of injury were found to be predominately high speed motor vehicle crashes followed by falls and pedestrians being struck [Ochsner et al. 1989].

Aortic rupture is thought to occur either from traction or shear forces generated between relatively mobile portions of the vessel and points of fixation or, secondly, due to direct compression over the vertebral column or, thirdly, caused by an excessive sudden increase of intraluminal pressure. Aortic rupture after thorax compression is shown schematically in Figure 5.5. Additionally, Viano (1983) reported that the inertial loading of the blood-filled heart can cause the heart to displace in the thoracic cavity and thus stretch points of attachment of the aortic arch, such as the superior arteries or the ligamentum arteriosum. This may occur if the heart is displaced vertically, laterally, or obliquely. Further, Viano considered the possibility of aortic laceration in combination with hyperextension of the neck in high speed loading (Figure 5.6). As for the site of thoracic aortic injury, it was found that the region of the aortic isthmus, just distal to the origin of the left subclavia artery, is most vulnerable. It accounts for 90% of
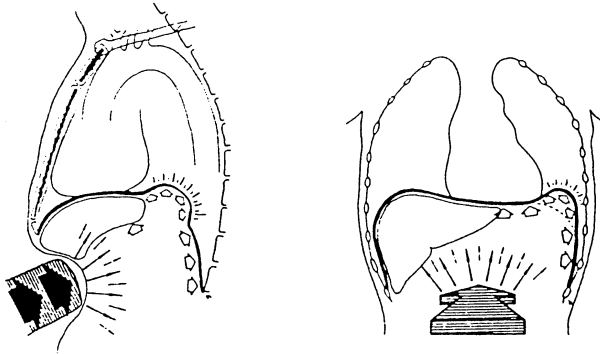
Fig. 5.7 Laceration of the diaphragm due to blunt impact on the abdomen [adapted from Viano 1990].

such injury [Creasy et al. 1997].

Other injuries of thoracic organs include rupture of the oesophagus and laceration of the diaphragm. The latter possibly results in a hernia. However, as Figure 5.7 shows, a laceration of the diaphragm is most probably a consequence from blunt impact to the abdomen (see chapter 6).

5.3 Biomechanical response

Many biomechanical tests have been performed under controlled laboratory conditions to measure the biomechanical response of the human thorax in terms of accelerations, forces, deformations and pressures. In particular cadaver tests, extensively conducted in the 1970s, were performed to obtain details of resulting injury to the body after impact. The data was then used to develop frontal and side impact dummies as well as to develop injury criteria. Furthermore the data was used to establish and validate mathematical models of the thorax.

In terms of test conditions, mainly pendulum and sled tests were used. Additionally, quasi-static tests – some with volunteers – to determine the stiffness of the thorax were performed.

5.3.1 Frontal loading

To investigate the biomechanical response of the thorax to frontal loading, extensive test series were performed. Human cadavers were impacted in pendulum tests using a 6-inch-diameter rigid pendulum (Figure 5.8).
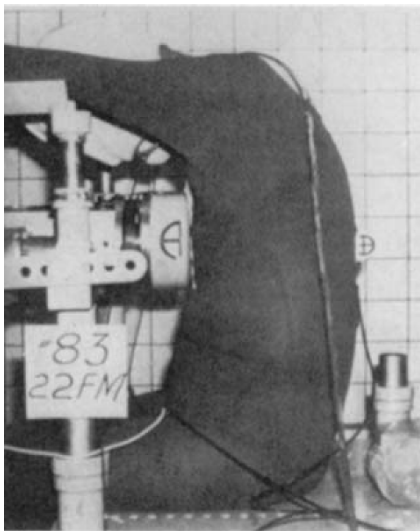
Fig. 5.8 Cadaver test using an impactor to apply load on the sternum [from Kroell et al. 1971].
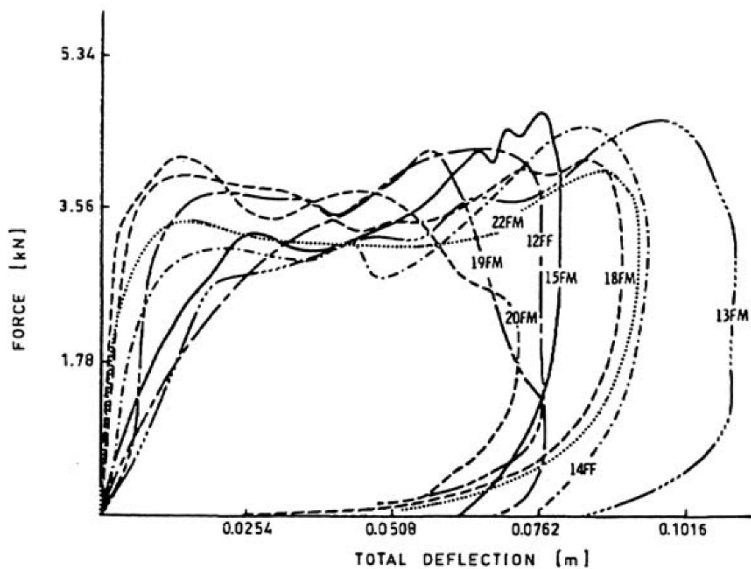
Fig. 5.9 Force-deflection characteristics of the thorax in frontal impact [from Kroell et al. 1974].

Measuring the deflection of the sternum, force-deflection characteristics for the thorax were determined [e.g. Kroell et al. 1971, 1974, Stalnaker and Mohan 1974]. Figure 5.9. shows a representative force-deflection curve as obtained from such experiments. The hysteresis curve can be divided into a loading and an unloading phase. The loading phase is characterised by an

initial rapid rise which is mainly due to the viscous properties of the thorax, and a plateau region that is also due to a viscous response. At maximum deflection, the impactor and the test subject are moving at a common velocity. The forces measured at this point are due to inertial forces caused by whole-body acceleration, and the elastic forces due to tissue compression. The unloading phase of the curve represents the unloading of the compressed tissues and follows the elastic non-linear unloading of the thorax. Analysing the relationship between the force plateau and impactor velocity, it was found that the force plateau increases with impactor velocity except for impactors with low mass but high velocity, which do not at all exhibit a force plateau. Furthermore, it was shown that lower impactor masses resulted in lower deflections [Lobdell et al. 1973].

Based on such cadaver tests, force-deflection corridors for different combinations of impactor mass and velocity were developed which are used for performance requirements for dummies.

In addition to dynamic pendulum tests focusing on sternal impacts, quasi-static tests have been performed. Since three-point belts and airbags are more frequently used today, lower rate loading has become more important in frontal impact. Distributed loading to the ribs due to airbags as well as rib and clavicle loading due to the shoulder belt make quasi-static thorax loading data also necessary. Performing such tests, the sternum of volunteers or cadavers is loaded with a plate with the subject's back against a rigid structure. The applied load and the anterior-posterior deflection of the thorax are recorded. Reviewing the data available, Melvin et al. (1985) concluded that for deflections of up to 41 mm the thorax has an approximate linear stiffness of 26.3 N/mm, and for deflections greater than 76 mm, the stiffness increases to 120 N/mm. However, the results are influenced by the individual physique of the test subject and differ remarkably for different conditions of the test subject such as embalmed and unembalmed cadavers and relaxed and tensed volunteers. In Figure 5.10 results obtained by Lobdell et al. (1973) are presented, clearly indicating the difference between a relaxed and a tensed volunteer. The fact that the stiffness of the thorax is increased in a tensed state can be regarded as beneficial in terms of injury tolerance.

The influence of wearing a seat belt, particularly a diagonal shoulder belt, on the occupant loading has been investigated since the late 1970s. It was observed that the thorax is more vulnerable to injury under the more concentrated belt loading. Injury due to belt loading appeared to be caused by thorax compression. More recently, studies analysed the effect of advanced belt systems that incorporate, for example, load limiters or pretensioners. From an analysis of accident data, Bendjellal et al. (1997)
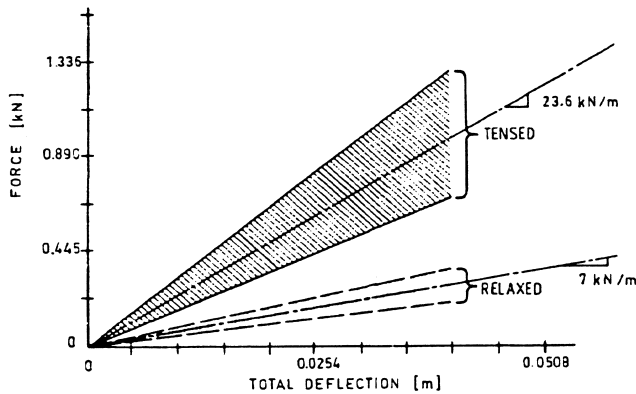
Fig. 5.10 Results from quasi-static volunteer tests showing the influence of a tensed and relaxed state of the thorax [from Lobdell 1973].

concluded that the shoulder belt force should be limited to 4 kN. Also Foret-Bruno et al. (1998) suggested a belt load limitation of 4 kN combined with a specially designed airbag system. They estimated that 95 % of AIS3+ thorax injuries could be prevented in frontal impacts.

Additionally, the loads to the thorax that are applied by a deploying airbag were assessed in several studies [e.g. Cavanaugh 2002]. In general, injury was related to the internal airbag pressure. When at any time during the deployment process the available volume of the airbag is smaller than the gas volume generated, high forces on the subject can arise. If an occupant is for instance “out of position”, i.e. if the subject is in the path of the airbag, a load on the subject’s thorax due to airbag pressure occurs. Figure 5.11 illustrates such loading caused by punch-out forces. Here the punch-out interaction is due to the proximity of the chest to the airbag system. Punch-out also occurs due to contact with the airbag module restricting the normal deployment for the airbag as it breaks out of the module, and begins to unfold. In contrast, membrane loading (Figure 5.11) occurs later in the deployment process after the airbag has left the module but with the occupant nonetheless being too close to the airbag. Consequently, the airbag wraps around the occupant, particularly in the head/neck and chest region. In addition to thorax injuries, head injuries, e.g. basilar fractures, are observed [McElhaney et al. 2002]. It appeared that the harmonisation of the airbag and the belt system is an important factor to maximise the benefits from such systems. From analysing accidents with deployment of the frontal airbag, Otte (1995) concluded that the protective effect of a three-point belt system was sufficient up to a delta-v of 35 km/h

to 40 km/h. For a higher delta-v, a supplementary airbag system should be activated. In contrast, Kallieris et al. (1995) reported that bag-like compression to the thorax is favourable, as the forces are distributed more evenly. Performing frontal impact cadaver tests (at 48 km/h impact velocity), they recommended to achieve overall restraint of the occupant by a belt, but that thorax injury mitigation should be aimed at using airbag systems. The combined thoracic index CTI was developed to assess both airbag and belt loading in crash tests (see chapter 5.4.5). Various tolerance values for frontal loading are presented in Table 5.2.

Based on experimental results, a mathematical model describing the behaviour of the thorax in frontal impact was developed [Lobdell et al. 1973]. The lumped-mass model utilises springs, masses and dampers (Figure 5.12) and the model's force-deflection response was tuned to match the low and high velocity corridors determined experimentally [Kroell et al. 1971, 1974]. In the meantime the model has been modified by various researchers and validated with additional test results. It is still used within the scope of dummy thorax design studies although finite element models are becoming more sophisticated (see chapter 2).
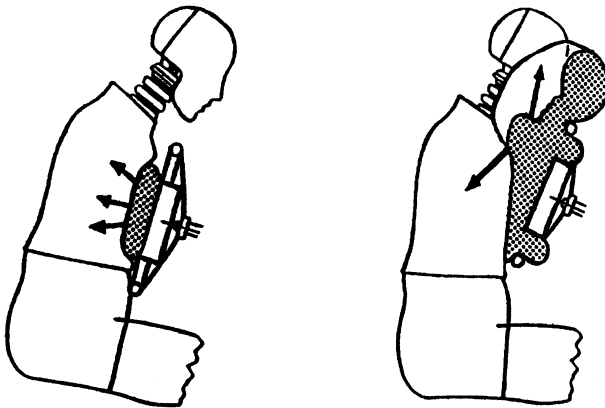
Fig. 5.11 Mechanisms for airbag inflation-induced injury. Punch-out loading mechanism causing pressure on the thorax (left) and membrane loading mechanism resulting in pressure to the thorax and the head-neck complex (right) [from Melvin and Mertz 2002].

Table 5.2 Frontal impact tolerances of the thorax.

tolerance level	injury level	reference
force:		
3.3 kN to sternum	minor injury	Patrick et al. (1969)
8.8 kN to chest and shoulders	minor injury	Patrick et al. (1969)
acceleration:		
60 g	3ms value for Hybrid III	FMVSS 208 (old version)
deflection:		
58 mm	no rib fracture	Stalnaker and Mohan (1974)
52 mm	limit for Hybrid III (5%)	FMVSS 208
63 mm	limit for Hybrid III (50%)	FMVSS 208
compression:		
20 %	onset of rib fracture	Kroell et al. (1971,1974)
40 %	flail chest	Kroell et al. (1971,1974)
VC_{max}:		
1.0 m/s	25 % probability of AIS ≥4	Viano and Lau (1985)
1.3 m/s	50 % probability of AIS ≥4	Viano and Lau (1985)
Combined Thoracic Index CTI:		
$A_{max}/60g+D_{max}/76mm$	50 % probability of AIS >3 in cadavers	Kleinberger et al. (1998)
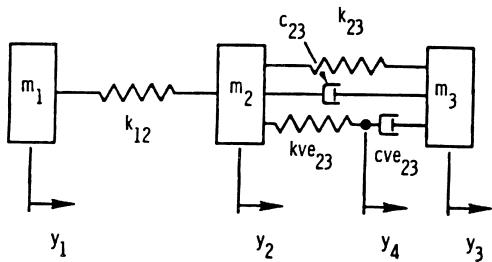
Fig. 5.12 Viscous thorax model [from Lobdell 1973].

5.3.2 Lateral loading

To investigate the biomechanical response of the human body in side impact, the same methods used to analyse frontal loading were applied. Cadaveric studies were also the method of choice to address the force-deflection characteristics of the thorax due to lateral impact. As a result of such impactor tests, hysteresis curves were presented that were similar to those obtained for frontal loading except for the fact that no or less apparent force plateau regions were determined. Furthermore, it was shown that the resistance of the thorax to lateral impact is smaller than to frontal loading. The arm of the test subject also has an effect on the test results depending on its position during impact. The arm can partly or completely be placed between the impacting mass and the thorax or it can be raised. Cesari et al. (1981) demonstrated this influence in a test series with cadavers. They concluded that the arm can have a protective effect when positioned between the striking object and the thorax.

In addition to the impactor tests, so-called drop tests were performed to analyse the force-deflection characteristics of the struck-side half-thorax. Cadavers were dropped from a height of 1 m to 3 m onto an unpadded or padded force plate [e.g. Stalnaker et al. 1979, Tarriere et al. 1979]. Results are summarised in Table 5.3. As a further result of these studies, a corridor for the development of a side impact dummy was proposed.

To further investigate side impacts, sled tests were conducted at the University of Heidelberg [Kallieris et al. 1981]. A seat with a low friction coefficient was mounted on a sled. The sled was suddenly decelerated from a specified velocity so that the test subjects (cadavers) sitting on the seat slid across the seat and impacted a padded or unpadded wall. The accelerations of the ribs, the sternum and the thoracic vertebrae were measured. It was noticed that besides the accelerations, physical parameters of the test subjects had a significant influence on the injury outcome. Consequently, the Thoracic Trauma Index (TTI) was proposed [Eppinger et al. 1984], which includes among others an age factor (see also section 5.4). Nowadays, side impact dummies allow to measure upper and lower rib accelerations so that the TTI can be calculated to assess side impact crash worthiness.

Table 5.3 Lateral impact tolerances of the thorax.

tolerance level	injury level	reference
force:		
7.4 kN	AIS0	Tarriere et al. (1979)
10.2 kN	AIS3	Tarriere et al. (1979)
5.5 kN	25 % probability of AIS ≥4	Viano (1989)
acceleration:		
T8-Y 45.2 g	25 % probability of AIS ≥4	Viano (1989)
T12-Y 31.6 g	25 % probability of AIS ≥4	Viano (1989)
60 g	25 % probability of AIS ≥4	Cavanaugh et al. (1993)
TTI(d):		
TTI(d) 85 g	Max. in SID dummy for 4-door cars	FMVSS 214
TTI(d) 90 g	Max. in SID dummy for 2-door cars	FMVSS 214
TTI 145 g	25 % probability of AIS ≥4	Cavanaugh et al. (1993)
TTI 151 g	25 % probability of AIS ≥4	Pintar et al. (1997)
compression to half thorax:		
35 %	AIS3	Stalnaker et al. (1979), Tarriere et al. (1979)
33 %	25 % probability of AIS ≥4	Cavanaugh et al. (1993)
compression to whole thorax:		
38.4 %	25 % probability of AIS ≥4	Viano (1989)
VC_{max} to half thorax:		
0.85 m/s	25 % probability of AIS ≥4	Cavanaugh et al. (1993)
VC_{max} to whole thorax:		
1.0 m/s	50 % probability of AIS ≥3	Viano (1989)
1.47 m/s	25% probability of AIS ≥4	Viano (1989)

5.4 Injury tolerances and criteria

As described in the previous sections, injuries to the thorax occur due to compression, viscous or inertial loading or combinations thereof. Using different kinds of experiments, the biomechanical response of the thorax in terms of tolerance values for various load cases was determined. Furthermore, injury criteria were developed to relate a certain loading of the thorax with an according injury risk. This chapter presents the most commonly used thoracic injury criteria and tolerance thresholds (Tables 5.2 and 5.3).

5.4.1 Acceleration and force

Early attempts to quantify thoracic loading focused on acceleration. As of today, the human tolerance for severe thorax injuries is considered as peak spinal acceleration sustained for 3ms or longer not to exceed 60 g in a frontal impact. This value is also embodied in FMVSS 208 to assess frontal impact crash worthiness. For lateral impact, different thresholds are proposed (see Table 5.2).

Closely related to acceleration is the definition of force tolerance values. Assuming an effective thorax mass of 30 kg, a force limit of 17.6 kN corresponds to the 60 g acceleration level. However, cadaver tests by Patrick et al. (1969) observed minor skeletal injuries already at 3.3 kN for impacts to the sternum and 8.0 kN for distributed loads to the shoulders and the thorax. These results show that the reliability of a single acceleration or force criterion as a general injury parameter for thoracic injuries is rather limited. Neither of the two criteria takes the viscous nature of the thorax into account. Consequently, more complex criteria were developed to obtain a better correlation with experimental results.

5.4.2 Thoracic Trauma Index (TTI)

The Thorax Trauma Index is an injury criterion for the thorax in the case of a side impact. It assumes that the occurrence of injuries is related to the mean of the maximum lateral acceleration experienced by the struck side rib cage and the lower thoracic spine. Furthermore, the TTI takes into account the weight and the age of the test subject and thus combines information on the kinematics with parameters of the subject's individual physique. The TTI (dimension [g]) is defined as follows:

$$TTI = 1{,}4AGE + 0{,}5(RIB_y + T12_y)(M/M_{std}) \qquad (5.1)$$

with AGE being the age of the test subject (in years); RIB_y [g] represents the maximum of the absolute value of the lateral acceleration of the 4th and 8th rib on the struck side; $T12_y$ [g] gives the maximum of the absolute value of the lateral acceleration of the 12th thoracic vertebra; M denotes the subject's mass [kg] and M_{std} refers to a standard mass of 75 kg.

When using a 50th percentile Hybrid III dummy to perform crash tests, a different version of the TTI, called the TTI(d), can be calculated. To obtain TTI(d) values, the age related term in equation 5.1 is omitted and the mass ratio becomes 1.0. It is important to note that the acceleration signals needed to determine the TTI and TTI(d), respectively, have to be preprocessed, i.e. filtered and sampled, according to a prescribed procedure (defined in FMVSS 214 and SAE J1727).

To relate TTI values to thoracic injuries, a large number of cadaver tests were performed [e.g. Kallieris et al. 1981] and injury risk functions were established statistically. Hence, the TTI reflects a statistical correlation rather than a biomechanical one. It cannot directly be related to any injury mechanism involved.

5.4.3 Compression Criterion (C)

Analysing blunt impact tests, Kroell at al. (1971, 1974) concluded that the maximum thorax compression correlated well with AIS while force and acceleration did not. Defining compression (C) as the chest deformation divided by the thickness of the thorax the following relationship was established:

$$AIS = -3{,}78 + 19{,}56C \qquad (5.2)$$

Thus measuring 92 mm thorax deflection for the 230 mm chest of the 50th percentile male results in a compression C of 40 % and predicts AIS4. 30 % compression lead to AIS2. Performing statistical analysis of the injury risk shows that in frontal impact a thorax compression of 35 % results in a 25 % probability of severe injuries rated AIS4 or higher. FMVSS 208 allows a maximum 76 mm deflection for the 50th percentile Hybrid III dummy in frontal impact.

5.4.4 Viscous Criterion (VC)

The viscous criterion (velocity of compression), also called the soft tissue

criterion, is an injury criterion for the chest area taking into account that soft tissue injury is compression-dependent and rate-dependent. The VC value [m/s] is the maximum of the momentary product of the thorax deformation speed and the thorax deformation. Both quantities are determined by measuring the rib deflection (side impact) or the chest deflection (frontal impact). Hence:

$$VC = V(t) \times C(t) = \frac{d[D(t)]}{dt} \times \frac{D(t)}{b} \quad (5.3)$$

where $V(t)$ [m/s] is the velocity of the deformation calculated by differentiation of the deformation $D(t)$, and $C(t)$ denotes for the instantaneous compression function which is defined as the ratio of the deformation $D(t)$ and the initial torso thickness b. Details on how the deformation data must be filtered is given in ECE R94 for side impact and SAE J1727 for frontal impact, respectively. Often the maximum VC, VC_{max}, which was found to correlate well with the risk of thoracic injuries [Viano and Lau 1985], is reported. Using the Lobdell model (see section 5.3.1), a relationship between the VC and energy absorbed in the thorax can be established. As for the critical values, both ECE R95 (lateral impact) and ECE R94 (frontal impact) require the VC to be less or equal to 1.0 m/s.

5.4.5 Combined Thoracic Index (CTI)

The Combined Thoracic Index represents an injury criterion for the chest area in case of frontal impact [Kleinberger et al. 1998]. Combining compression and acceleration responses, the CTI particularly addresses both airbag and belt loading. The CTI is defined as the evaluated 3 ms value from the resultant acceleration of the spine and the deflection of the chest. The calculation of the CTI value is based on the following equation:

$$CTI = \frac{A_{max}}{A_{int}} + \frac{D_{max}}{D_{int}} \quad (5.4)$$

where

A_{max} = 3 ms value (single peak) of the resultant acceleration of the spine [g]

A_{int} = critical 3 ms intercept value [g]

D_{max} = deflection of the chest [mm]

D_{int} = critical intercept value for deflection [mm]

Intercept values are defined for different dummy types. For the 50th percentile Hybrid III, for example, they read 85 g for A_{int} and 102 mm for D_{int}.

The combined compression and acceleration criterion accounts for the differences in loading of the thorax by belt versus airbag systems. It is based on the assumption that, for a given load, a belt system would apply greater pressure along its contact area than an airbag system, which has a larger contact area. With the combined belt/airbag system, the predominant loading could be a line load, i.e. the load from the belt is larger than the one from the airbag, or a distributed load in the opposite case. The CTI is meant to reflect the whole range of possible loading scenarios between these two extremes. While the maximum thorax acceleration is a measure of the magnitude of total forces applied to the torso in proportion to its mass, the thorax deflection is an indicator of the belt loading. The greater the deflection per unit of acceleration, the more the relative contribution of the belt system [Cavanaugh 2002].

The CTI was developed based on cadaver tests and correlated to the AIS by logistic regression analysis. To date, the CTI was included in FMVSS 208, where details on data acquisition and the different intercept values are given.

5.4.6 Other criteria

The Rib Deflection Criterion (RDC) is the criterion for the deflection of the ribs, expressed in mm, in a side impact collision. According to ECE R95 the RDC shall be less than or equal 42 mm (side impact dummy).

ThCC (or TCC) is the abbreviation for Thoracic Compression Criterion. ThCC is the criterion of the compression of the thorax in frontal impact between the sternum and the spine and is determined using the absolute value of the thorax compression, expressed in mm. Today a maximum threshold value of 50 mm is defined in ECE R94.

5.5 Thoracic injuries in sports

With respect to thoracic injuries specifically related to sports not much is reported in the literature. The aforementioned descriptions of injuries and injury mechanisms also apply for traumatic injuries in sports. Additionally overuse injuries might occur, e.g. in form of rib stress fractures in elite rowers [Karlson 1998], but this seems to be a rather rare phenomenon.

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6 Abdominal Injuries

The abdominal cavity is a vulnerable region of the human body. In general, trauma to the abdomen is caused by blunt impact or by penetration. In automotive accidents, blunt impact is frequently observed although the injury might not be apparent initially. Nonetheless, it may be life-threatening. In side impacts, for example, it was found that more than one fifth of all severe injuries, i.e. AIS $\geq$ 4, were abdominal [Rouhana and Foster 1985].

In terms of investigating the biomechanical response of the abdomen, experimental studies turned out to be particularly difficult to perform and the results obtained are not easy to interpret. Thus, we still lack sufficient knowledge of injury mechanisms and appropriate injury predictors. This lack of knowledge is also evident in human surrogates used in crash testing (see chapter 2). An excellent review on abdominal injuries was presented by Rouhana (2002).

6.1 Anatomy of the abdomen

Cranially, the abdominal cavity is bounded by the diaphragm and caudally by the pelvic bones and the muscles attached to it. The lumbar vertebral column, which itself is usually not considered as part of the abdomen, forms, together with the sacrum and the pelvis, the posterior boundary of the abdomen. Anteriorly and laterally the upper abdomen is bound by the lower rib cage. The lower abdomen is surrounded anteriorly and laterally by musculature. Because of the ribs, the upper abdominal region, sometimes called the “hard thorax” [Eppinger et al. 1982], shows a different behaviour with respect to impact response and tolerance than the lower abdomen. The presence of the lower ribs (although not directly attached to the sternum, see chapter 5) comes particularly into effect in rear-end and side impacts. For frontal impact, however, it appears that organs

directly in front of the vertebral column are at higher risk of being compressed than organs lateral to the spine.

The abdominal cavity hosts several organs that are generally divided into "solid" and "hollow" organs. The main characteristic to divide the organs into these two groups is the gross density of an organ (not the tissue density). Solid organs like the liver, spleen, pancreas, kidneys, ovaries and adrenal glands have a higher density than hollow organs such as the stomach, large and small intestines, bladder and uterus. The lesser density of the hollow organs is due to the presence of a relatively large cavity within the organ itself. Those cavities are, for example, filled with "air" or digestive matter. The solid organs, in contrast, contain fluid-filled vessels and therefore exhibit a higher density.

Major blood vessels of the abdomen are the abdominal aorta, the inferior vena cava, the hip artery (arteria iliaca communis) and the hip vein (vena iliaca communis). The abdominal aorta and the vena cava enter the abdomen from the cranial side through separate openings in the diaphragm. Figure 6.1 illustrates the abdominal organs in situ. With respect to the biomechanical response of the abdomen on traumatic impact, it is important to note that the organs inside the abdominal cavity inhere a relatively high degree of mobility. They are neither rigidly fixed to the abdominal wall nor to each other. Partly they are embedded in fat (e.g. the kidneys) or are tethered by folds of the peritoneum (e.g. the intestines). The peritoneum is a serous membrane that covers the inner abdominal walls and surrounds each organ. As this membrane is smooth and moist, it acts as a lubricant and thus also adds to the mobility of the organs. Consequently, the abdominal organs can adjust to different postures such as sitting or standing. Furthermore, the position of the liver, for instance, changes during respiration, because it moves with the diaphragm. This mobility has therefore a great influence on the biomechanical response and, of course, also on the outcome of experimental studies examining the injury mechanism.

In summary from an anatomical point of view, possible injury mechanisms seem to be dependent on the complex structure of the abdomen together with physical properties like the density, structure and the material within the organs.

6.2 Injury mechanisms

Due to the complex structure of the abdomen, there are many factors influencing the location, likeliness and severity of a blunt impact. First of
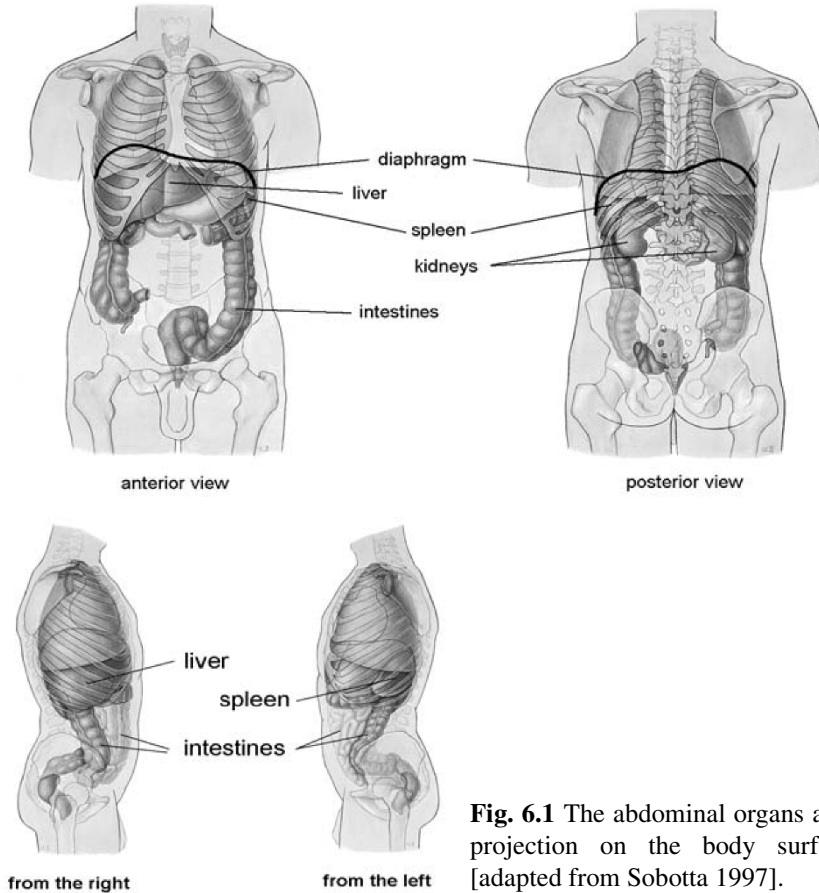
Fig. 6.1 The abdominal organs as a projection on the body surface [adapted from Sobotta 1997].

all, anatomy suggests that the position of an organ contributes to its injury risk. Organs that are located anterior of the vertebral column are, in case of a frontal impact, more likely to be compressed against the spine than those lying lateral. Additionally, the upper abdomen is in part covered by the lower rib cage, which also has an protective effect in frontal impact. Obviously, the non-symmetric organisation of the abdominal organs accounts for different injury risk depending on the impact direction. If the abdomen is struck from the right side, liver injury is more probable than if struck from the left. As the lung, the liver can experience a central rupture where the tissue around the damaged part is not altered.
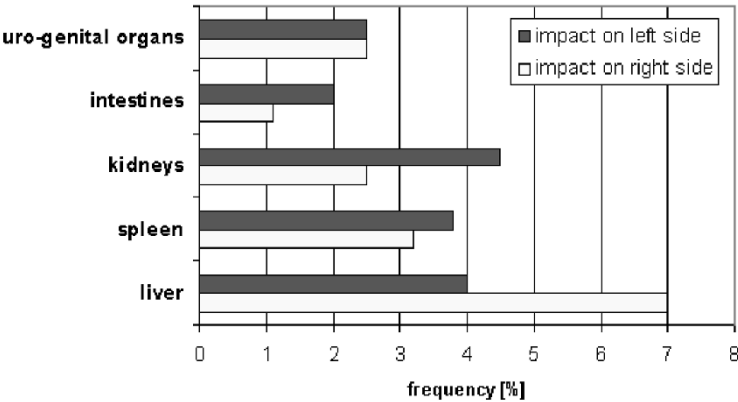
Fig. 6.2 Frequency of AIS > 3 abdominal injury for different organs due to lateral impact on the right or left side, respectively [adapted from Rouhana et al. 1985].

Table 6.1 Abdominal injuries [AAAM 2005].

AIS code	description
1	skin, muscle: contusion (hematoma)
2	spleen or liver contusion (<50% surface area)
3	major kidney contusion spleen: rupture
4	abdominal aorta: minor laceration kidney/liver: rupture
5	kidney: total destruction of organ and its vascular system
6	hepatic avulsion (total separation of all vascular attachments)

Figure 6.2 presents the injury frequency for different organs due to lateral impact on the right and the left side, respectively. Examples of possible abdominal injuries and their classification according to the Abbreviated Injury Scale (AIS) are given in Table 6.1.

In automotive crashes, the vehicle interior offers several contact areas that strongly influence the injury outcome when hit. Possible contact points include the steering assembly, the side door, the arm rest, the dashboard and the glove compartment, whereas unbelted occupants are, of course, at higher risk to contact such structures than belted ones.

Also the structure of the organs themselves is important in terms of abdominal injury. Solid organs were found to be injured more often than hollow organs. Furthermore, the pathological state of an organ can have a marked influence on the injury tolerance due to changes of the material properties (e.g. stiffness). Among other medical conditions, previous surgery resulting in adhesions inside the abdominal cavity are also suspected to predispose the subject to injury. The age of a patient was shown to affect the injury outcome in blunt trauma, with children and the elderly being at higher risk. Particularly for children anatomical reasons have to be considered as, for example, the abdominal region is proportionally larger than in an adult, the liver is less protected by the rib cage and thus at higher risk. However, Khaewpong et al. (1995) analysed cases of restrained children that sustained abdominal injury, finding nearly 90% of these injuries were associated with contact with the restraint system. In fact, all of those children injured were using the restraint system either incorrectly, inappropriately, or both incorrectly and inappropriately.

6.3 Testing the biomechanical response

In the same way as for other regions of the human body, the biomechanical response of the abdomen is addressed in experimental studies utilising cadavers or animals. Several studies used impactors that completely covered the abdomen (in front as well as in lateral impact). However, the overall response curves obtained do not account for the non-homogenous nature of the abdomen. The location of impact (e.g. left side or right side) and the posture of the object during impact play important roles.

The test conditions chosen do, of course, also influence the outcome of the experiments. Analysing the biomechanical response of the abdomen, the so-called “fixed back” condition is often used. By fixing the back of the object during impact the influence of the spine is eliminated.
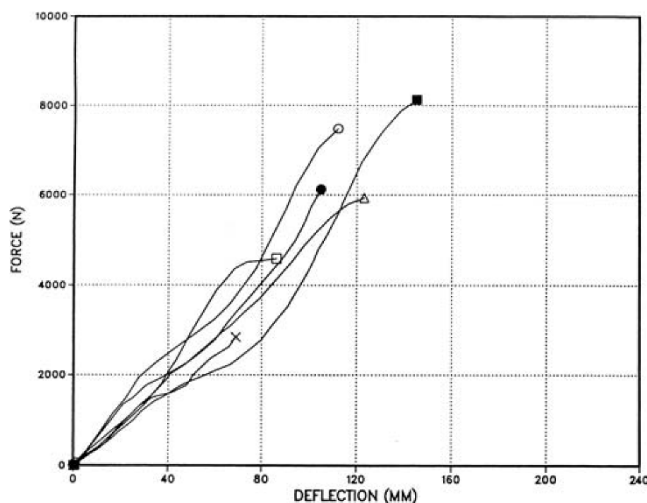
Fig. 6.3 Force-deflection characteristics for the lower abdomen obtained in frontal impact tests using a rigid impactor [adapted from Nusholtz et al. 1988].

In terms of experimental procedure it turned out to be quite difficult to accurately determine the deflection of the abdomen. Mostly high speed videos were evaluated, sometimes determining the intrusion with respect to a fixed point (e.g. the spine) or measuring the compression relative to the other side of the object. Due to the poor quality of most of these videos, many studies were not able to determine the deflection and therefore reported the force history only. Others, as pointed out by Rouhana (2002), presented results that are not reliable.

Today, force-deflection curves obtained from cadaver tests are available for the lower abdomen in frontal impact [Cavanaugh 1986, Nusholtz et al. 1988] (Figure 6.3). Furthermore experiments with anaesthetised pigs were performed for frontal impact condition [Miller 1989]. For the upper abdomen in frontal impact, it is suggested to used the same data as for the lower abdomen until better data is available [Rouhana 2002]. With respect to lateral impact, several cadaveric studies were presented performing sled tests, pendulum tests and drop tests. Drop tests especially addressed the impact on an arm rest, dropping the cadaver from a certain height on the arm rest [e.g. Walfisch et al. 1980]. Force-time curves instead of force-deflection curves were presented (see above).

For blunt impact on kidneys, Schmitt et al. (2006a,b) carried out pendulum tests on human as well as porcine kidneys. It was shown that the kidney tissue failure is a predominately energy driven phenomenon. The
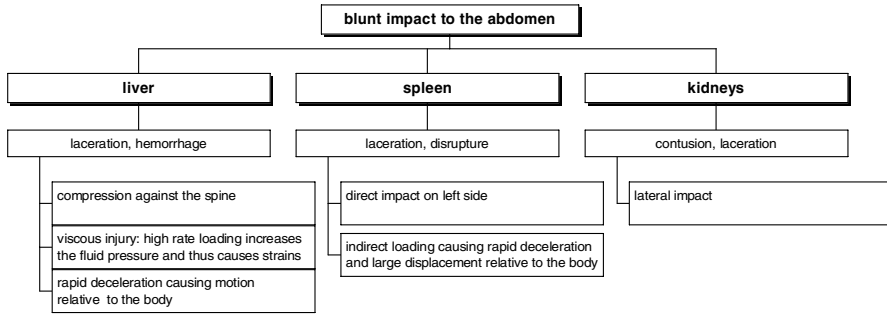
Fig. 6.4 Possible injuries and injury mechanisms for different abdominal organs.

visco-elastic material properties were described and force-deformation characteristics are provided.

In Figure 6.4 possible injury mechanisms and the resulting injury are presented for the three organs injured most often.

6.4 Injury tolerance

Several experimental studies were performed to find ways to quantify the biomechanical response of the abdomen to impact and consequently to develop suitable injury criteria. The strength of various mechanical parameters as predictors of abdominal injury was investigated. Although various possibilities were addressed, not many conclusions in terms of tolerance thresholds could be drawn. This, again, reflects the complex nature of the abdomen and the difficulties associated with the performance of adequate experiments. Consequently, further basic research is needed to elucidate the injury mechanism and appropriate injury tolerance levels. This paragraph summarises the most important attempts to quantify predictors for abdominal injury.

In general it is assumed that the force on an occupant should well correlate with injury outcome. Experiments with anaesthetised rabbits under condition of lateral impact, for example, confirmed this hypothesis by showing that the peak force correlated well with the probability of AIS $\geq$ 3 renal injury [Rouhana et al. 1986]. However, it did not correlate with the probability of hepatic injury. Testing swine, Miller (1989) found that the peak force correlated well with the likeliness of AIS $\geq$ 3 and AIS $\geq$ 4 lower

abdominal injury in belt loading. While no threshold values could be determined from the animal experiments, the results from human cadaver studies proposed a maximum tolerable force value of 4.4 kN [Talantikite et al. 1993].

In contrast, acceleration was found not to be a good measure for abdominal injury [Rouhana 2002]. A major problem is the actual measurement of the acceleration. Typically the acceleration was obtained from accelerometers attached to the spine and the rib cage, respectively. Attaching accelerometers to those structures, they basically record the whole body acceleration. Thus a good correlation to injuries of the abdomen cannot necessarily be expected.

Bearing in mind that the solid organs of the abdomen are “fluid-filled”, a rate dependent behaviour was suspected. In fact, several studies including one by Mertz and Weber (1982), who performed tests on pigs, found a strong influence of the rate of abdominal compression and injury. Analogously to the thoracic impact, a good correlation to the injury severity was obtained when calculating the product of the maximum impact velocity V and the maximum abdominal compression C [e.g. Rouhana et al. 1984, Stalnaker and Ulman 1985]. It was shown that for very low loading velocities (e.g. seat belt loading), the maximum compression was a better predictor of abdominal injury. For high loading velocities (e.g. airbag loading), the maximum velocity was a better injury predictor. For compressions and velocities in between, the product of V*C was found to be a better predictor than either the maximum velocity or the maximum compression separately. In addition, it was also shown that the product of maximum force F and maximum abdominal compression C correlate well with the probability of AIS ≥ 4 injury [Rouhana 1987].

In recent work on blunt impact of kidneys an impact energy threshold of 4 J, or a corresponding strain energy density of 25 kJ/m^3, were found to cause moderate to severe renal injury [Schmitt et al. 2006a,b].

6.4.1 Injury criteria

To date, only the European regulation for side impact testing (ECE R95) proposes a threshold level for abdominal loading. The abdominal peak force (APF) as determined by use of the EuroSID dummy is required to be less than or equal to 2.5 kN internal force (which is equivalent to 4.5 kN external force).

6.5 Influence of seat belt use

Within the scope of abdominal injuries, the influence of the seat belt is often discussed. Since the 1960s the so-called “seat belt syndrome” has been reported in the literature. Blunt impact to the abdomen is assumed to be caused by the seat belt either because of submarining and/or because of misplacement of the belt. Both submarining as well as misplacement are primarily related to lap belts or the lap part of 3-point belts, respectively. Submarining occurs in crashes with high change of velocity (Δv) when the occupant’s pelvis manages to slip underneath the lap belt such that the lap belt then loads the abdomen. Hence, the structure of the seat strongly influences the probability of submarining. To prevent this movement, the seat base cushion often exhibits a wedge like shape at the frontal end, or anti-sliding-airbags might be introduced.

If the lap belt is not positioned properly (misplacement), i.e. if the belt is placed above the pelvis, it also loads the abdomen instead of the more stable pelvis. The correct placement of the belt is particularly crucial for pregnant women to ensure that the fetus is not exposed to high loading. Nonetheless, it is important to state that pregnant women should definitely wear the belt. Special devices to enable a correct path of the belt are commercially available.

Besides these two possible reasons for abdominal injury, the overall effectiveness of the seat belt was proven in many studies, showing that unbelted occupants are twice as likely to sustain fatal injuries than belted occupants [e.g. Langwieder et al. 1990, Lane 1994, Rouhana 2002]. Furthermore, Langwieder et al. (1990) reported that up to 90% of seat belt associated injuries are AIS1 injuries. Nonetheless, it can be suspected that the pattern of injury changes due to seat belt use. While the belt effectively reduces head, neck and thorax injury, it might possibly be responsible for more frequent but minor abdominal injuries if worn incorrectly [Harms et al. 1987].

6.6 Abdominal injuries in sports

Blunt or penetrating trauma to the abdomen is rare, but the literature reports some cases of hernia and groin injury. The "sports hernia" can occur in athletes who participate in sports that require repetitive twisting and turning at speed (e.g. ice hockey, soccer, tennis, field hockey). However, in many cases, an actual hernia is not seen. Several theories exist in the literature

regarding the causes of the sports hernia most of which implicate an overuse syndrome. Hip abduction, adduction, and flexion-extension with the resultant pelvic motion produce a shearing force across the pubic symphysis, leading to stress on the inguinal wall musculature perpendicular to the fibers of the fascia and muscle. Pull from the adductor musculature against a fixed lower extremity can cause significant shear forces across the hemipelvis. Subsequent attenuation or tearing of the transversalis fascia or conjoined tendon has been suggested as the source of pain [Anderson et al. 2001].

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7 Injuries of the Pelvis and the Lower Extremities

Injury to the lower extremity play a major role in sports and have emerged as the most frequent non-minor injury resulting from frontal vehicle crashes, often resulting in long-term impairment [Håland et al. 1998, Crandall 2001].

In this chapter injuries of the lower limbs are discussed. A short review of the anatomy is followed by a description of possible failure mechanisms and resulting injuries. The biomechanical response is analysed and criteria developed to predict injuries of the lower limbs are presented. A further section deals specifically with injuries sustained in sports.

7.1 Anatomy of the lower limbs

The lower limbs are commonly divided into pelvis, thigh, knee, lower leg, ankle and foot (Figure 7.1). The pelvis which links the lower extremities to the spine is a ring of bones basically composed of four bones: two hipbones form the side and front walls while the sacrum and the coccyx form the rear wall (Figure 7.2). Mechanically the pelvis represents the only load path to transmit the weight of the torso to the ground. Hence, its structure is quite massive. The hipbones consist of three fused bones (ilium, ischium, pubis) and also host the acetabulum, a cup-shaped articular cavity forming one part of the hip joint. The pubic bone and the pubic symphysis, i.e. the joint connecting the right and the left pubic bone, form the frontal part of the pelvis. Particularly the thinner, frame-like parts of these pubic bones, the superior and inferior pubic rami, are often subjected to injury. At the rear wall of the pelvis, the sacrum is a fusion of the sacral vertebrae with sacral nerves (e.g. the sciatic nerve) that arise from the spinal cord passing the sacrum. Major blood vessel are also located near the sacrum and the coccyx.

Fig. 7.1 Anatomy of the lower limbs [adapted from Sobotta 1997].
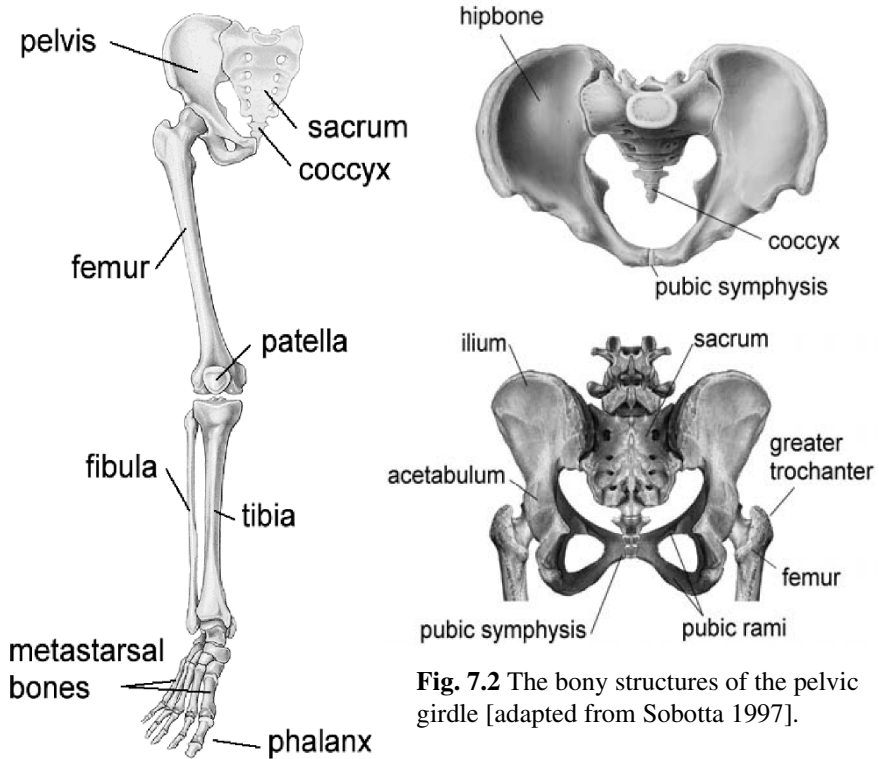
Fig. 7.2 The bony structures of the pelvic girdle [adapted from Sobotta 1997].

As for the orientation of the pelvis, Figure 7.3 illustrates the position of the bony structures in different postures. It is obvious that injury resulting for example from impact to the knee depends on the posture at the moment of impact.

Differences between the male and female pelvis are also apparent, but will not be discussed here. Although the shape and the mechanical properties of the male and female bones are slightly different, no differences in injury mechanisms concerning automotive impacts are reported in the literature.

The femur is the long bone of the thigh and is proximally connected by the hip joint to the pelvis and distally linked to the knee. The different regions and landmarks of the femur are shown in Figure 7.4. Two bones, the tibia and the fibula, form the lower leg between the knee and the ankle. The knee is the joint that connects the femur and the lower leg (Figure 7.1). It is an anatomically dense area involving several muscles, tendons,
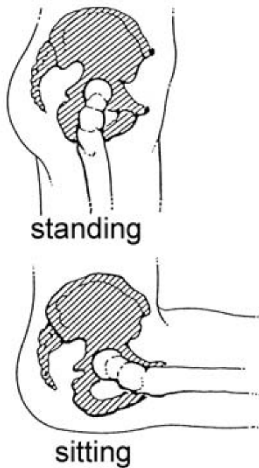
Fig. 7.3 Orientation of the hip in different postures: standing (top) and sitting (bottom) [adapted from Kramer 1998].
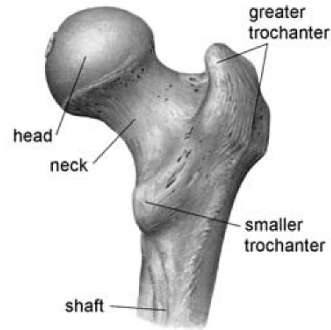
Fig. 7.4 Regions and landmarks of the femur [adapted from Sobotta 1997].
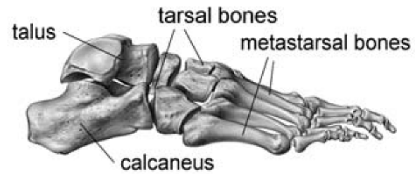
Fig. 7.5 Bony structures of the foot [adapted from Sobotta 1997].

ligaments and menisci. Moreover, vulnerable structures of the knee like the patella are often subjected to direct impact. A strong musculature which can create considerable forces and thus may influence the injury mechanisms (see section 7.2.2) surrounds the legs.

Finally the foot is adjoined to the lower leg. A foot consists of several bones: calcaneus and talus are located at the proximal end, the metastarsal bones and phalanx at the dorsal end (Figures 7.1 and 7.5).

7.2 Injury mechanisms

Regarding the pelvis and the lower extremities, fractures are the most common injuries sustained in accidents. Such fractures result from sports accidents or falls, respectively, rather than from automotive accidents. Hip fractures, for example, that are often caused by falls, particularly concerning the elderly, are a major concern in public health. Worldwide,
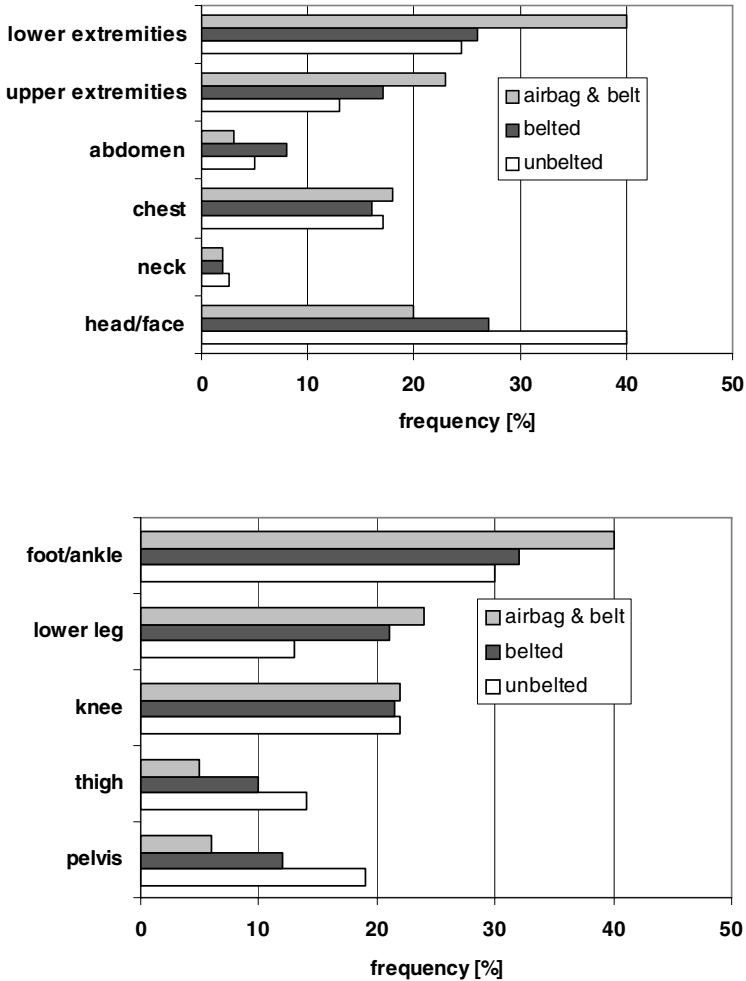
Fig. 7.6 Distribution of AIS ≥ 2 injuries in frontal impact by body regions (top) and by lower limb regions (bottom) for unbelted occupants, belted occupants and occupants wearing the belt plus having the airbag deployed. [adapted from Crandall 2001].

there are approximately 1.7 million people suffering from hip fractures yearly [Kannus et al. 1999]. In contrast, pelvis fractures sustained in automotive accidents are quite rare. They contribute to only about 1 % the total Injury Priority Rating (IPR) [King 2002]. Analysing frontal impacts of passenger cars, Kramer (1998) found that, while head injuries were the

most common injuries sustained by 35 % of all occupants, pelvis and hip injuries were present in only about 7 % of all cases. However, 25 % of the occupants had leg and foot injuries. Similar results were obtained when evaluating the NASS data base [Crandall 2001]. Moreover, checking for AIS ≥ 2 injuries (AIS, i.e. Abbreviated Injury Scale) in frontal impact, a strong influence of the restraint systems on the likeliness of injuries of the lower extremities was apparent (Figure 7.6). It was observed that the percentage of lower limb injuries is about twice as high than that for head injuries when the occupant is belted and the vehicle is equipped with airbags. The analysis of the different regions of the lower extremities showed that the feet and the ankles are at highest risk for AIS ≥ 2. Additionally, Morris et al. (2006) found that, based on UK accident data, lower extremity AIS ≥ 2 injuries are by far the most costly injuries and account for some 43% of injury costs in both front and struck-side crashes.

Due to the fact that the pelvis and the proximal femur are often simultaneously injured, such injuries are commonly referred to as hip injury whereas the word "hip" does neither describe a particular anatomical structure that is injured nor is "hip injury" related to a particular injury mechanism. In the strict sense, the hip is the bony structure around the hip joint (femur head, pelvis, acetabulum). However, fractures involving the proximal part of the femur are commonly called hip fractures as well.

Generally, fractures are either open or closed. While the skin and soft tissues overlaying the fracture are intact in closed fractures, the bone is exposed to direct outside contamination in open fractures. Further characteristics to classify fractures include the position of the fractured segments (displaced/undisplaced), the location of the fracture along the bone (intraarticular, metaphyseal, diaphyseal) and others [see e.g. Levine 2002]. Concerning fracture of long bones in general and with particular respect to the bony structures of the legs, fracture patterns are differentiated depending on the loading condition that caused the fracture, i.e. the injury mechanism. There are four possible types of fracture mechanisms: direct loading, indirect loading, repetitive loading and penetration. In motor vehicle accidents direct and indirect loading are the most frequent types. If in a frontal collision the knee of an occupant hits, for example, the dashboard, direct loading can cause fracture of the patella, indirect loading may lead to fractures of the femur shaft or the acetabulum (Figure 7.7). Different fracture patterns that can arise from direct and indirect loading, respectively, are presented in Figure 7.8. Like other injuries, injuries of the pelvis and the lower extremities are categorised in the AIS (Table 7.1).

In the following sections possible injuries and their injury mechanisms are discussed; they are limited, however, to impact related injures only.
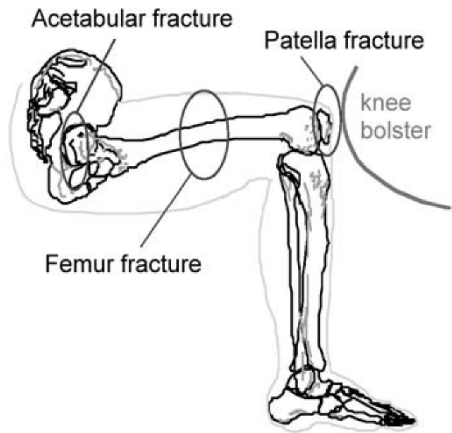
Fig. 7.7 Possible fractures due to impact to the knee [adapted from Crandall 2001].

Table 7.1 Examples of AIS rated injuries of the pelvis and the lower extremities [AAAM 2005].

AIS code	description
1	ankle, hip: sprain, contusion
2	patella, tibia, fibula, calcaneus, metatarsal: fracture pelvis: fracture (closed, undisplaced) toe: amputation, crush hip, knee dislocation muscles, tendons: laceration (rupture, tear, avulsion)
3	femur: fracture pelvis: fracture (open, displaced) traumatic amputation below knee
4	pelvis: “open book” fracture traumatic amputation above knee
5	pelvis: substantial deformation with associated vascular disruption and blood loss > 20% by volume
6	-
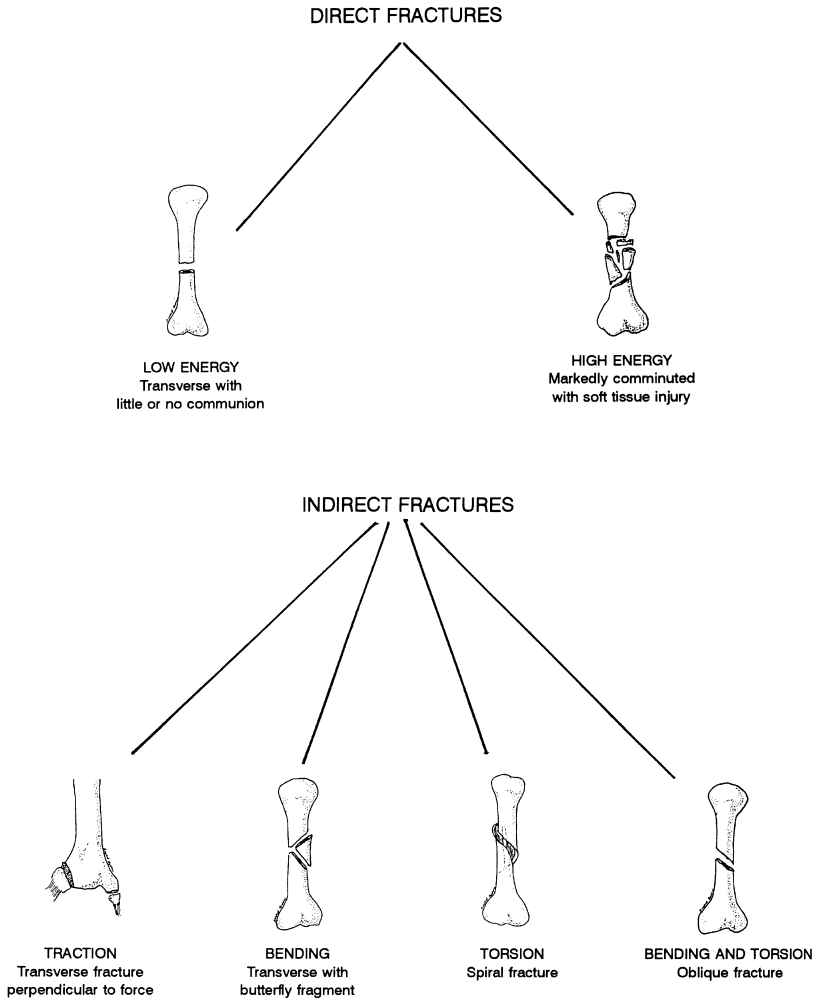
Fig. 7.8 Types of fracture according to loading [from Levine 2002]. It should be noted that bending fractures can also occur directly, i.e. they are also a form of direct fracture.

7.2.1 Injuries of the pelvis and the proximal femur

Injuries to the pelvis are categorised clinically as isolated fracture of the pelvic ring, multiple fractures of the pelvic ring, sacrum fracture and associated injuries. An isolated fracture of the pelvic ring consists of a single fracture around the pelvic ring (for example of the pubic rami or the ilium). Pubic rami fractures are frequently observed in lateral impacts when the greater trochanter is hit. The pelvic ring remains stable after isolated fractures, i.e. significant displacement of the fractured segments is not found. This is different if multiple fractures occur. Here the pelvic ring becomes unstable enabling large displacement of the fractured segments. Uro-genital injuries can arise in combination with multiple pelvic fractures.

Sacrum fracture occurs in extensive pelvic injuries, fracturing usually across the foramina or in the vicinity of the holes through which the sacral nerves pass. Obviously, these nerves are also at danger in case of such injury. Moreover additional injury, especially haemorrhage, can be associated with pelvic fractures. Excessive bleeding from the large blood vessel in the pelvic wall as well as from the fractured surfaces themselves can be life-threatening (even if proper surgery is applied).

From a biomechanical point of view, the underlying mechanisms of pelvic fracture are either compression, vertical shear or a combination thereof. Compression of the pelvis can further be differentiated into lateral and frontal (i.e. anterior-posterior) compression. Figure 7.9 illustrates the possible locations of fracture in case of lateral compression. If an anterior-to-posterior directed force compresses the pelvis centrally, so-called straddle fractures occur, i.e. multiple fractures of the pubic rami. Anterior-posterior (a-p) compression with impact forces on the right and left iliac crest can result in a hinge or “open book” fracture (Figure 7.9). Thus in a-p compression the pelvic diameter is acutely increased causing tensile forces to act on tissue hosted by the pelvis and possibly violating ligaments. If the
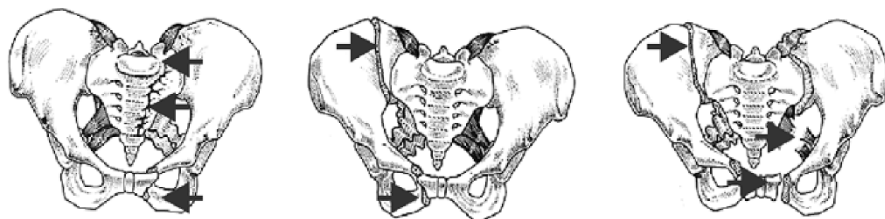
Fig. 7.9 Possible locations for fracture arising from lateral compression [adapted from Vetter 2000].
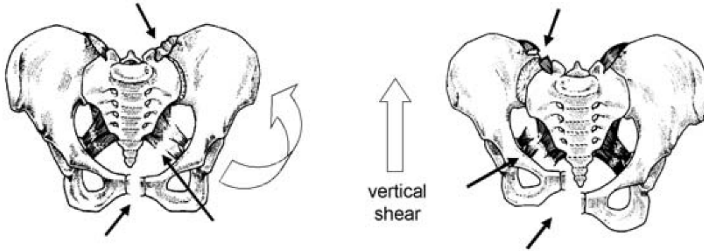
Fig. 7.10 Fracture and ligament rupture due to vertical shear forces (right). "Open book" fracture (left).

Fig. 7.11 Left: Dislocation of the right hip. Right: Dislocation and acetabulum fracture of the left hip after lateral impact. The head of the femur broke through the medial wall of the acetabulum.

pelvis is subjected to vertical loading, shear can cause fracture as well as rupture of ligaments (Figure 7.10).

Pelvic fractures used to be predominant in pedestrian collisions up to the early 1990s. When the pedestrian was struck by a car laterally on the pelvis, fractures of the pubic rami occurred (prevailing on the non-struck side). This type of injury has nearly vanished in today's pedestrian collisions. Due to changes in the front shapes and the front structures of recent cars, pedestrian kinematics seem to be different such that pelvic fractures are prevented [Otte 2002].

As described above, the hip is frequently injured in falls. Such lateral loading of the hip commonly causes the femur neck to fracture. Femoral neck fractures in geriatric patients tend to be very low energy injuries [Levine 2002]. In automotive accidents, however, fracture of the proximal femur is rare. Lateral impact is more likely to cause fracture of the pubic rami instead. Luxation and dislocation of the hip joint (Figure 7.11),

possibly in combination with fracture of the acetabulum, can also result from lateral impact. In general, the hip joint is dislocated when the hip is flexed and adducted and simultaneously loaded along the femur in rearward direction. These conditions apply, for instance, to occupants experiencing a frontal collision while sitting with their legs crossed.

7.2.2 Leg, knee and foot injury

Addressing the injury mechanism for femur fractures sustained in frontal automotive collisions by interaction with the dashboard, it was found that axial compression accounts for the vast majority of fractures (62%), followed by bending (24%), torsion and shear (each 5%) [Crandall 2001]. In addition to the mechanical loading conditions, the fracture pattern of femur shaft fractures is influenced by the fact that the femur is not perfectly straight, i.e. it is slightly curved so that the convex side faces forward. This plays a role especially in indirect loading of the femur (Figure 7.7).

Impact to the knee can cause the patella to fracture (direct loading of the patella). Likewise indirect loading of the patella can occur from strong muscle contraction (quadriceps) on the partially flexed knee, possibly resulting in patella fracture [Levine 2002]. Unlike the hip joint which has a stable ball-and-socket joint (section 7.2.1), the bone anatomy of the knee imparts little support to the joint's stability. This makes the knee ligaments prone to injury (see also section 7.5). When the knee is bent and an object forcefully hits the tibia backwards, the posterior cruciate ligament can tear (this injury is commonly called "dashboard injury"). In lateral impacts (e.g. pedestrian impacts) also collateral ligament rupture is seen. Generally complete dislocation of the knee joint can result in tears of the major four knee ligaments.

In analogy to femur fractures, tibia fractures may be caused through direct or indirect loading of the leg. Tibia fractures are the most common fractures in long bones [Crandall 2001], and because the tibia lies subcutaneously (i.e. the bone is only covered by the skin), such fractures are often open fractures. Most fractures occur between the mid-shaft region and the distal third of the tibia, where the smallest cross-section and the smallest cross-sectional moment of inertia are found. As for the impact conditions, Crandall (2001) reports that for tibia as well as for fibula injuries, axial loading and direct impact are equally probable. If both the tibia and the fibula are fractured, stability depends on the level of the fractures, i.e. the fracture is more unstable if both bones are broken on the same shaft level. A more severe type of tibia fracture is the tibia plateau
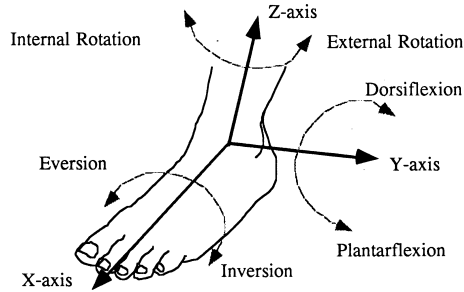
Fig. 7.12 Anatomical motions of hindfoot joints [adapted from Crandall et al. 1996].

fracture, i.e. a fracture of the tibia including a disrupture of the articular surface of the knee joint. Tibial plateau fractures make up for only 1% of all fractures seen clinically, but account for 10% of all below-knee AIS ≥ 2 injuries sustained in vehicle crashes [Crandall 2001]. Mechanisms resulting in tibia plateau fracture include direct impact to the knee, axial loading and axial loading plus hyperextension of the knee. Axial loading might occur due to simultaneous floorboard intrusion and knee entrapment (e.g. if the knee gets caught under the dashboard).

The injury mechanisms of ankle and foot injuries are closely related to the possible motion range of the ankle and the hindfoot (Figure 7.12). According to Crandall (2001), ankle injuries sustained in frontal automotive collisions are primarily due to axial load (58%), inversion (15%) and eversion (11%). Only 5% resulted from direct contact. Metatarsal injuries, however, are solely caused by direct impact and 100% of all calcaneus injuries are due to axial loading. Pure axial loading can also result in talus fractures and so-called pylon fractures, i.e. intra-articular fractures of the distal tibia involving the calcaneus, talus and the distal tibia, for instance by driving the talus into the tibia. Dorsiflexion (e.g. forced in crashes with toepan intrusion) and entrapment of the knee can increase axial forces.

Inversion and eversion account for the vast majority of malleolar injuries, especially ankle fractures, making the ankle the most frequently injured major joint of the human body (see also section 7.5). Furthermore, the rate of rotation, the orientation of the ankle and individual factors of the occupant like age or pre-existing damage were found to influence the likeliness of malleolar injury [Crandall 2001].

Foot injuries, particularly metatarsal fractures, that were sustained in automotive accidents, result mainly from contact to the foot pedals. If, in addition, the heel strikes the floorboard, local loading to the calcaneus

occurs, possibly leading to its fracture [Levine 2002].

Investigating the effect of preload in the Achilles tendon with respect to ankle injuries, cadaver experiments showed that a 2 kN Achilles preload affects pylon fracture by decreasing the external force needed to cause this fracture [Kitagawa et al. 1998a]. Achilles preload also leads to tension fractures of the calcaneus. Additionally, injury mechanisms are influenced considerably by the musculature of the lower extremities. Cappon et al. (1999) report the posterior muscles (plantarflexors) to be able to generate more than 3.5 kN and for the anterior muscles (dorsiflexors) 1.4 kN were determined [Petit et al. 1998]. Likewise, in pre-impact bracing muscles can contribute more than 2.8 kN to the load of the lower extremities [Crandall et al. 1995]. Sled testing and computer simulations showed that muscle tension largely increases the axial forces, increases the effective mass and stiffness of the leg, accounts for redistribution of the stress within the bones, and alters the kinematics with less forward excursion and small joint rotation [Crandall et al. 1995]. However, to gain a better understanding of the injury mechanisms of lower limb injuries, more studies addressing the role of muscle force are necessary.

7.3 Impact tolerance of the pelvis and the lower extremities

The response to mechanical loading of isolated bones such as the femur, the tibia or the fibula was measured from experiments similar to those used in material testing (e.g. bending and tension tests). Table 7.2 summarises such

Table 7.2 Mechanical strength (average values) of the bones of the lower limbs as reported by Levine (2002).

	femur		tibia		fibula	
	male	female	male	female	male	female
Torque [Nm]	175	136	89	56	9	10
Bending [kN]	3.92	2.58	3.36	2.24	0.44	0.30
Average maximum moment [Nm]	310	180	207	124	27	17
Long axis compression [kN]	7.72	7.11	10.36	7.49	0.60	0.48

results.

To determine the impact response of the pelvis cadaver tests subjecting the test objects to vertical, frontal and lateral loading were performed. Because of the use of very different test set-ups (e.g. rigid versus padded impactors) and different test procedures (e.g. instrumentation and posture of the test object), the results obtained for each load case are difficult to compare. As in some tests hardly any fractures were produced, the evaluation of the results is not at all conclusive.

To mimic the loading conditions of a automotive frontal collisions, Nusholtz et al. (1982) performed pendulum tests impacting a sitting cadaver at the knee. No pelvic or hip fractures were observed up to 37 kN impact force. Brun-Cassan et al. (1982) conducted whole-body impacts on unrestrained cadavers with peak knee forces ranging from 3.7 to 11.4 kN. No pelvic fractures were found, except for one case where the right patella and the iliac crest fractured at a knee load of 8.8 kN. Despite the differences found in the cadaver tests, a maximum of 10 kN for axial loading of the femur is included in FMVSS 208.

Concerning lateral impact, tests results reported in the literature are also diverse. In summary, it appears that neither the peak pelvic acceleration nor the peak pelvic deformation correlate well with the probability of pelvic fracture. However, Viano et al. (1989) found that the ratio of pelvic deformation to pelvic width (presented in percent compression) was a reliable measure for fractures of the pubic rami. 27 % pelvic compression corresponded to a 25 % probability of serious injury. This finding was confirmed by Cavanaugh et al. (1990), who also performed lateral impact on cadavers. A slightly higher percentage of tolerable compression (32.6 %) was determined to account for a 25 % probability of fracture of the pubic rami.

Also the (visco-elastic) material properties and the failure thresholds of soft tissue of the lower extremities were characterized by various experimental studies leading to a large spread of results. Since for tendons and ligaments the ultimate load is related to the cross-sectional area such variations are to be expected. The average ultimate tensile stress of tendons and ligaments as described in the literature ranges from 50 to 100 MPa. Depending on the experimental setup threshold values for failure of the knee ligaments due to strain are reported from 7 to 40 % [e.g. Butler et al. 1986, Kerrigan et al. 2003, Arnoux et al. 2004]. Figure 7.13 presents some typical stress-strain-curves for knee ligaments. Dynamic tests performed on cadaveric knees suggest a 50% risk of the collateral ligament injury to be associated with an applied bending moment of 117 to 134 Nm [Ivarsson et al. 2003]. However, in several other studies a wide range of failure
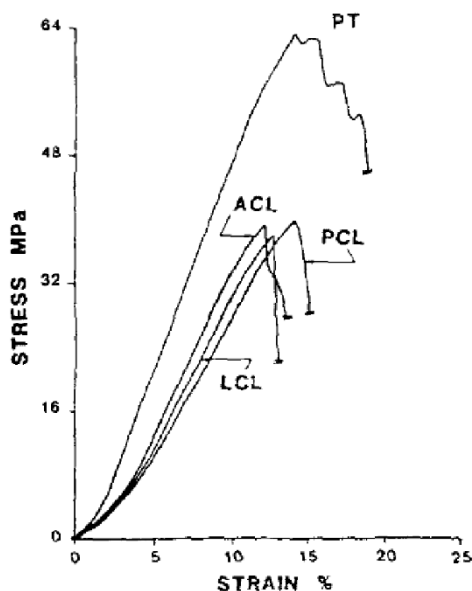
Fig. 7.13 Stress-strain curves for the patella tendon (PT), the anterior (ACL) and posterior (PCL) cruciate ligaments and the lateral collateral ligament (LCL) fascicle-bone units [adapted from Butler et al. 1986].

thresholds is reported so that general conclusions can not be drawn.

With respect to the lower leg and the foot, various cadaver and volunteer experiments were conducted. The static response to axial loading was analysed for both cadavers and volunteers [Hirsch and White 1965]. Force-deflection response was obtained. A comparison between the cadaver and volunteer results suggests that there is no difference between the compressive foot and ankle stiffness of the living human and the cadaver specimens [Crandall et al. 1996]. Dynamic response in axial loading of the tibia and the ankle was also determined from cadaver studies [e.g. Petit et al. 1996, McMaster et al. 2000]. Figure 7.14 illustrates a test set-up using a pendulum to impact the foot and the tibia. Reported failure loads for tibia fracture are for example 7.8 kN [Begeman and Prasad 1990] and 8.0 kN [Yoganadan et al. 1996], 7.3 kN are given for pylon fractures [Kitagawa et al. 1998b], and 8.1 kN were found for calcaneal fractures [Begeman and Prasad 1990].

For dorsiflexion/plantarflexion the biomechanical response was investigated in static and dynamic tests. From cadaveric studies it emerged that after 45 deg of forced dorsiflexion, there is a 50% probability of ankle injury [Levine 2002]. Rudd et al. (2004) concluded that an ankle joint moment of 59 Nm represents a 25% risk of ankle fracture in dorsiflexion for a 50%ile male. Parenteau et al. (1998), who performed quasi-static experiments, reported injury at 47.0±5.3 deg and 36.2±14.8 Nm in dorsi-
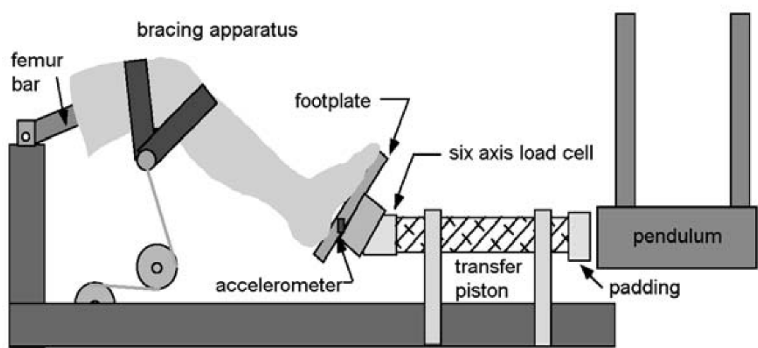
Fig. 7.14 Test apparatus to perform dynamic impact experiments [adapted from Crandall 2001].
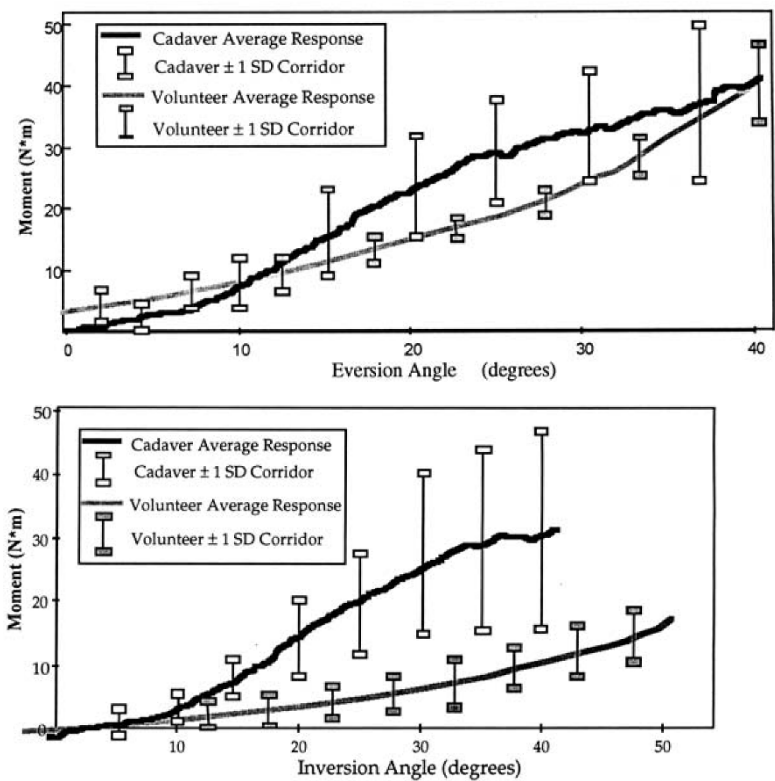
Fig. 7.15 Response corridors for eversion (top) and inversion (bottom) as obtained from quasi-static volunteer and cadaver tests [from Crandall et al. 1996].

flexion, and at 68.7±5.9 deg and 36.7±2.5 Nm in plantarflexion. Dynamic experiments resulted in injury at 138 Nm but exhibited a large variability of ±50 Nm [Begeman and Prasad 1990].

Likewise eversion and inversion of the foot was analysed statically and dynamically. For quasi-static loading, Parenteau et al. (1998), for example, found a failure threshold of approximately 34 Nm and 48 Nm for inversion and eversion, respectively. Figure 7.15 presents the according response corridors as reported by Crandall et al. (1996).

Bearing in mind the additional influence from active and passive musculature, preload as well as individual factors like bone mineral density [McMaster et al. 2000], further experiments are necessary to improve the definition of impact tolerances of the lower limbs and thus to provide information for the development of accurate injury criteria, improved mathematical models and biofidelic dummies.

7.4 Injury criteria

To assess loading of the pelvis and the lower extremities in crash tests, few criteria are established in existing regulations. This is not surprising, as the total number of specimens tested in the underlying investigations concerning loading mechanisms is still small. However, additional criteria are proposed but not yet included into test standards.

7.4.1 Compression force

To protect the hip-thigh-knee complex a maximum compression force of 10 kN for axial loading of the femur is defined in FMVSS 208.

The tibia compression force criterion (TCFC) as defined in ECE R94 determines the force axially transmitted to each tibia of a test dummy. To date, the maximum threshold value for TCFC is 8 kN.

7.4.2 Femur Force Criterion (FFC)

FFC as defined in ECE R94 assesses the compression force acting on the femur as well as the duration [ms] for which the force is applied. The FFC is determined by the compression force [kN] that is transmitted axially on each femur. Figure 7.16. presents the force limits that must not be exceeded in a test.
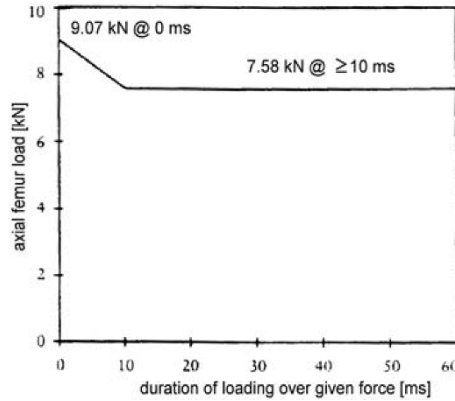
Fig. 7.16 Femur force criterion as defined in ECE R94.

7.4.3 Tibia Index (TI)

The Tibia Index (TI) involves the bending moments as well as the axial force in the tibia. The underlying idea of the TI is to prevent tibia shaft fractures. The TI is calculated according to the following equation:

$$TI = \frac{M}{M_{crit}} + \frac{F}{F_{crit}} \tag{7.1}$$

with M being the bending moment and F the compressive force. M_{crit} and F_{crit} represent critical intercept values and read 225 Nm and 35.9 kN, respectively, for the 50th percentile male. These critical values were obtained in static bending tests of the tibia [Yamada 1970]. The maximum TI measured at the top and bottom of each tibia shall not exceed 1.3 at either location (ECE R94). As a further restriction, a maximum compression force was defined, i.e. the maximum compression force measured has to be smaller than 8.0 kN. Using scaling techniques, according critical values were determined for a 5th percentile Hybrid III female and for the 95th percentile Hybrid III male dummy. The detailed evaluation procedure including the required filtering is described in ECE R94.

7.4.4 Other criteria

For side impact, the maximum strain on the pubic symphysis is taken as a measure for pelvic strain. The according criterion (cf. ECE R95) is called

pubic symphysis peak force (PSPF) and shall not exceed 6 kN.

A maximum tibial displacement of 15 mm to protect the knee ligaments is laid down in ECE R94 (frontal impact). Moreover, a maximum ankle and foot load of 7.5 kN to protect the hindfoot and the ankle is discussed.

7.5 Pelvic and lower extremity injuries in sports

With respect to sports injuries of the lower extremity, several extrinsic and intrinsic factors and their influence on the injury risk are discussed [see e.g. Murphy et al. 2002]. While it is generally accepted that the injury incidence is greater during competition than in training sessions, controversial results are presented for the influence of skill level, shoe type or ankle bracing. As for intrinsic factors there is strong evidence that previous injury, especially when followed by inadequate rehabilitation, increases the injury risk while a correlation between injury risk and, for example, limb dominance, fitness status, body size or flexibility is not generally established or depends strongly on the study design. This inconsistency in the literature reflects the large variability represented by the different sports and the individual athlete and makes it difficult, if not impossible, to systematically present injury mechanisms and injury thresholds.

Concerning the lower extremities, injury to the muscle is frequent. Blunt impact, i.e. caused by an impactor's knee, compresses the muscle and is the predominant mechanism for muscle contusion. In approximately 9% to 17% after a direct blow to a muscle myositis ossificans traumatica develops whereas the incidence is also thought to be related to the severity of injury. Myositis ossificans traumatica is a nonneoplastic proliferation of bone and cartilage in an area previously exposed to trauma and haematoma. Its origins, relationship to other forms of bone proliferation (after surgery, congenital), and treatment are less than clear [Beiner and Jokl, 2002].

Hamstring strains are common in sports that involve running or sprinting and jumping, but are also common in dancing and waterskiing. A problem with this injury is the high rate of re-injury. It is suspected that hamstring strains develop during the later part of a swing phase when the hamstrings are working to decelerate knee extension i.e the muscle develops tension while lengthening. Consequently, the hamstrings must change from functioning eccentrically to concentrically which is suggested to make the muscle vulnerable to injury [Peterson, Hölmich 2005].

The bony structures of the lower extremity can be subjected to various scenarios that result in direct or indirect fracture due to mechanical loads

exceeding the fracture threshold (see sections 7.2 and 7.3). Additionally, repeated, but sub-critical loading induces cumulation of microtrauma which can result in stress fractures (cf. chapter 9). Stress fractures of tibia, femur or the metastarsal bones are observed in long distance runners or ballet dancers.

The high stability of the hip joint ensures that dislocation and subluxation are infrequently observed in sports, though not impossible. Anterior dislocations are sometimes reported from high-energy collisions in skiing and contact sports. Pelvic fractures, however, are unusual injuries in athletes [Anderson et al. 2001]. The proximal femur is prone to fracture from direct loading (see section 7.2.1) as well as to overuse injury in terms of stress fractures. Under normal conditions, the downward bending moment (force on femoral head times length of femoral neck) induces tension stresses and strains in the superior aspect of the femoral neck. These are counteracted by contraction of the abductor muscles producing a compensatory compressive strain on the superior aspect of the femoral neck. Now, if the gluteus medius muscle is fatigued, this neutralizing effect is minimized and tensile strains are experienced by the superior aspect of the femoral neck [Egol et al. 1998]. Hence, in particular if loaded repeatedly stress fracture can occur.

The various structures of the knee are prone to injury from direct or indirect loading. As already mentioned in section 7.2.2 fracture of the patella can occur. Further patella disorders include patella tendon ruptures, patellofemoral pain syndrome or patellar tendiopathy (formerly known as "jumper's knee"). The majority of knee injuries sustained in sports, however, concern the ligaments and the menisci. Evaluating over 7000 knee injury records, Majewski et al. (2006) found that most injuries were related to the anterior cruciate ligament (ACL) (20.3%) followed by medial meniscus lesion (10.8%), lateral meniscus lesion (3.7%), medial collateral ligament (MCL) lesion (7.9%), lateral collateral ligament (LCL) lesion (1.1%), and posterior cruciate ligament (PCL) lesion (0.65%). The activities leading to most injuries were soccer (35%) and skiing (26%). LCL injury was associated with tennis and gymnastics, MCL with judo and skiing, ACL with handball and volleyball, PCL with handball, lateral meniscus lesions with gymnastics and dancing, and medial meniscus lesions with tennis and jogging.

The mechanical function of the menisci is related to weight bearing, shock absorption, stabilisation and rotational facilitation. Failure of the menisci generally involves shear and compression. Meniscus tears are caused by a body rotation around the fixed and weight bearing knee. This can occur either in a combination of flexion and rotation or extension and

rotation during weight bearing.

ACL rupture occurs most often in response to valgus loading in combination with external tibial rotation or to hyperextension with internal tibial rotation [Whiting and Zernicke, 1998]. The first mechanism can for instance be observed in a rugby or American football when the foot is on the ground, bearing weight, and another player contacts the lateral aspect of the knee increasing the valgus loading and rotation. Based on the second mechanism, ACL ruptures in basketball or gymnastics are recorded with the injury occurring after the athlete lands following a jump.

With respect to skiing, Hunter (1999) summarized three common injury mechanisms that result in ACL rupture: (1) valgus-external rotation (catching an inside edge and falling forward between the skis), (2) the boot-induced ACL injury (landing on the back of the ski with an extended knee, resulting in the boot forcing the tibia anterior as the front of the ski hits the ground) and (3) the phantom-foot phenomenon (falling backward between the skis, catching the inside edge of the downhill ski, driving the leg into forced internal rotation).

Injury of the PCL is less common, but occurs when the tibia is forced posterior relative to the femur (see section 7.2.2 "dashboard injury"). Figure 7.17 illustrates an example. Tearing of the PCL is also possible in a fall on the flexed knee pushing the tibia rearwards, if the knee is forced in flexion with the foot plantar-flexed or in rapid cutting on a minimally flexed knee [Whiting and Zernicke, 1998].

It is interesting to note that female athletes are at higher risk for knee injury than their male counterparts. This holds particularly true for non-contact ACL ruptures [Dugan 2005]. Several causes are discussed for this higher incidence rate including differences in anatomy and physiology (e.g. femoral notch dimensions, muscle strength, ligament dominance, sexual hormones) as well as dynamic neuromuscular imbalance. An extensive

Fig. 7.17 PCL tears if the tibia is forced rearwards relative to the femur [adapted from Peterson and Renström 2002].

review on this topic can be found in Dugan (2005).

Ankle sprain is among the most often reported injury in sports. According to Whiting and Zernicke (1998), the vast majority (85%) of ankle sprains result from inversion injuries (also called supination) typically due to rolling the ankle while the foot is in contact with ground. With respect to further injuries of the ankle, the foot and the Achilles tendon, the reader is referred to section 7.2.2. In addition overuse injuries to the foot (stress fracture) and the Achilles tendon (degeneration possibly even leading to rupture) occur and, similarly to the overuse injuries mention above, afflict mainly endurance athletes or military recruits (see chapter 9).

7.6 Prevention of lower extremity injuries

As already pointed out in previous chapters, the strategies to prevent injuries in sports are very much depending on the type of sports and the individual circumstances. Thus it is not surprising that the protective potential of several measures is discussioned controversially. The use of prophylactic knee braces [e.g. Najibi and Albright 2005] as well as the effect of shin guards to prevent tibia fractures [e.g. Francisco et al. 2000] are such examples.

In the automotive environment, in contrast, more agreement exists concerning countermeasures to reduce the number of injuries of the pelvis and the lower extremities sustained in automotive crashes. Knee bolsters that are located in the lower part of the dashboard were introduced to protect the knee from impact and also to provide an additional load path for the deceleration of the lower body in the absence of seat belts. To prevent knee ligament injury, it is important that the knee bolsters also contact the upper patella area in order to ensure that forces are induced axially to the femur. Likewise, knee airbags were developed to reduce loading on the knee in frontal collisions (Figure 7.18). Recently, first car models that are equipped with such a knee airbag system were made available.

To keep the occupant's knees and legs at a safe distance from the instrument panel, a system called anti-sliding-bag was presented [Autoliv 2003]. The anti-sliding bag is installed in the front edge of the seat cushion to prevent the occupant from sliding under the seat belt in a crash. Thus the system also prevents submarining (see chapter 6) and is therefore meant to also reduce the risk for injuries to the abdomen.

Furthermore, the structure of the passenger compartment can be designed such that intrusion and intrusion velocity, respectively, are

significantly reduced in the region of the lower limbs. With respect to foot injuries, a re-design of foot pedals is thought to reduce the loads transmitted. To help prevent debilitating leg and ankle injuries caused by deformation of the footwell, foot airbags to be placed underneath the carpet were developed. By lifting the heel in case of a frontal impact, dorsiflexion of the foot is reduced. Håland et al. (1998) showed that such airbags reduced the foot acceleration by up to 65%, the tibia force by up to 50% and the tibia index by 30% to 60%.
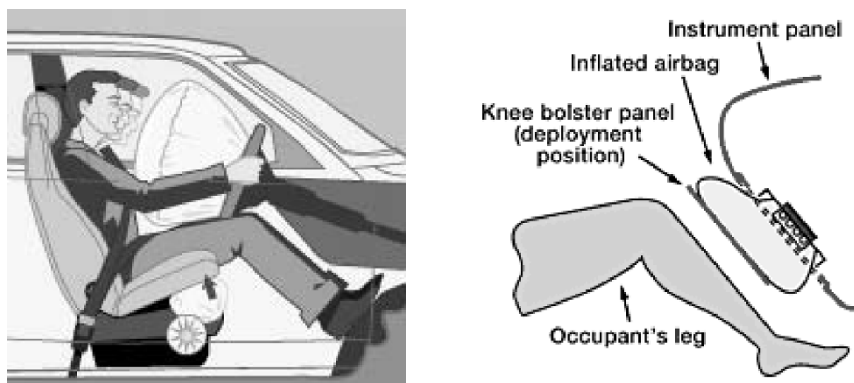
Fig. 7.18 Measures to prevent lower limb injuries. Knee airbag (right) and anti-sliding-bag (left). The anti-sliding-bag also prevents submarining (see chapter 6) [Autoliv 2003].

7.6.1 Pedestrian injury countermeasures

While pedestrian sensing and warning technologies are designed to prevent accidents, parallel research is aimed at helping to reduce injuries when an impact is unavoidable. One goal is to make a vehicle less likely to cause injury to pedestrians, for example by minimising the aggressiveness of the geometry and the structure of the vehicle front.

With respect to the geometry, recent studies show a strong influence of the injury outcome depending on the front shape. Evaluation of the data base of the Medical University of Hanover showed for example that pelvic and upper leg injuries are only prevalent in cars with rather sharp and high bonnet leading edges [Otte 1999, 2002]. This finding was corroborated by Snedeker et al. (2003) who performed numerical simulations finding that a car which exhibits a low bonnet leading edge height (<750mm), a large bonnet edge radius (>250mm), a moderate bumper lead (>150mm) and a sufficiently high bumper edge height (>490mm) would practically exclude the possibility of a pelvic/upper leg fracture in primary lateral pedestrian

impact at impact velocities up to 40 km/h. Furthermore it was shown that the degree of bonnet leading edge roundness has an important effect on the upper leg kinematics of pedestrian impact. There also appeared to be a direct relationship between bonnet leading edge sharpness and resulting mechanical stress in the acetabulum.

The closing speed of contact between the thigh and car bonnet is a critical factor determining injury risk. The closing speed is often not equivalent to the collision speed and may heavily depend on the radius of the bonnet leading edge [Snedeker et al. 2003].

With respect to other features of the front geometry, it has long been realised that bonnet ornaments can be dangerous in case of an impact. Therefore, today, bonnet ornaments have to be mounted such they can flap or rotate and (in Switzerland) the shape of the ornament must be designed according to the guidelines by the Swiss Federal Roads Authority [ASTRA 2002]. Furthermore, so-called bull bars were found to be causing injuries, especially when children are impacted. Thus the use of such devices is restricted in Switzerland [ASTRA 2002].

Another relevant aspect in pedestrian-vehicle collisions is the structure of the car front, particularly its deformation properties. The design of bumpers may be optimised to allow a higher degree of energy absorption. Several designs having been presented to date showed that bumpers incorporating multi-density foams and a structural undertray (secondary loadpath) to support the legs of a pedestrian, reduce impact forces and thus decrease the risk of leg and knee injuries. Moreover, the head lamp surrounding can be re-designed using a deformable housing which allows pushing the headlamps back into the car body to reduce the risk of injury.

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8 Injuries of the Upper Extremities

In automotive accidents, injuries to the upper extremities have probably received least of all attention compared to other body segments. This lack of interest can partly be explained by the fact that such injuries are generally not life-threatening. Nonetheless, they may cause long-term impairment associated with significant societal cost. Mainly the introduction of supplementary restraint systems, i.e. airbags, accounts for a recently regained interest in upper limb injuries. Due to their proximity to the upper limbs, side airbags were especially under scrutiny but also the effect of front airbags that deploy close to the hands and forearms is investigated. Furthermore, developments in airbag design, like depowered airbags, require current views to be constantly revised.

Although a few research groups have addressed impact response issues lately, the work by Messerer (1880) is still one of the most often quoted studies in this field. With regard to accident statistics, a similar lack of basic knowledge becomes evident. Only a few studies are available that focus on injuries of the upper extremities. Considering that the vehicle fleet is not yet equipped with (side) airbags in high numbers, any conclusions have to be drawn carefully. Hence, the analysis of upper limb injuries and related subjects is still an open field in which much progress can be expected over the next years.

In contrast to the automotive environment, injuries to the upper extremities are common in sports and therefore received considerable attention. Various studies addressed the kinematics of the upper extremities in different motion patterns like throwing, a golf swing or a tennis stroke. Many studies can also be found on the diagnosis and treatment of upper extremity sports injuries. With respect to injury mechanisms, however, many questions remain unanswered and concerning injury criteria and injury threshold levels basically no conclusive literature is available.
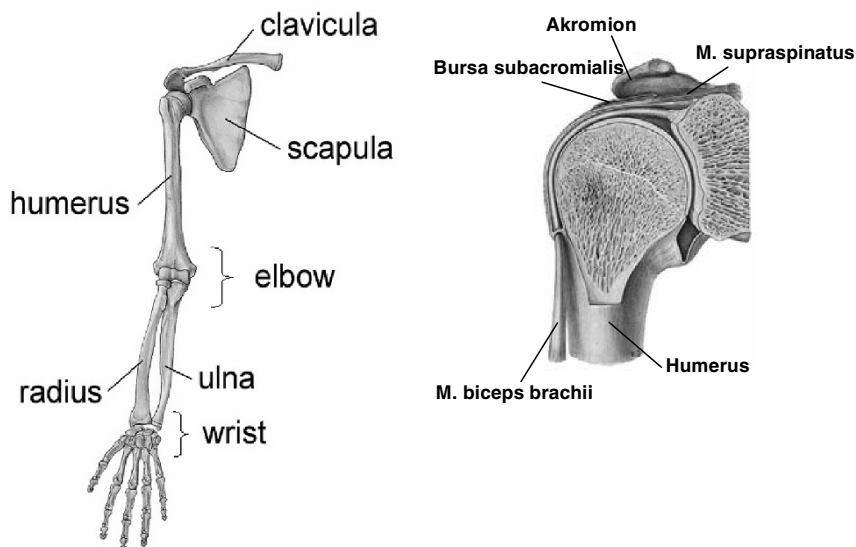
Fig. 8.1 Bones and joints of the upper extremities [adapted from Sobotta 1997].
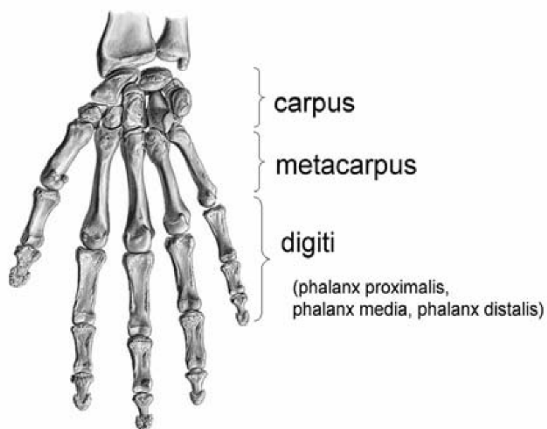
Fig. 8.2 Skeleton of the hand [adapted from Sobotta 1997].

8.1 Anatomy of the upper limbs

Generally, the upper extremities can be divided into four different parts: the shoulder (or shoulder girdle), the arm, the forearm and the hand. Figures 8.1 and 8.2 illustrate the corresponding bony structures.

The shoulder comprises scapula, clavicula and the joint articulations that

attach the upper extremities to the torso. The arm is formed by the humerus and is linked to the shoulder by the shoulder joint which is probably the most mobile joint in the human body. The movement of the clavicula and scapula allows translation of the shoulder in horizontal and frontal planes. Additionally rotations about the three anatomical axes are provided by the shoulder joint.

The elbow joint connects the arm to the forearm which consists of the ulna and the radius. A much simpler joint than the shoulder joint, the elbow joint allows flexion of the forearm towards the humerus, extension of the forearm away from the humerus and one half of the forearm pronation/supination rotations. These pronation/supination rotations are completed by the ulna rotating at the wrist. The wrist joint, finally, connects the forearm to the hand. Associated muscles and soft tissues complete the four parts of the upper extremities.

It should be noted that there are differences between the upper extremities of males and females. Within the scope of trauma-biomechanics, the most relevant differences are the mass and the bone mineral density. Both are lower in women with the bone mineral density also being further reduced with age.

8.2 Injury incidences and mechanisms

Injuries of the upper limbs focus on fractures of the long bones. Of course, the soft tissues and the muscles may also be injured (e.g. skin abrasions due to airbag contact [Duma et al. 2003], but these injuries play a minor role in the field of automotive accidents.

As for fractures, the classification presented in chapter 7 also applies. Most common are clavicula fractures which occur, for instance, in direct blows, through compression during lateral impact of the shoulder or in falls on the outstretched arm. A typical fracture of the ulna is the so-called nightstick fracture which is a diaphaseal ulna fracture unaccompanied by a radius fracture. It results from low-energy direct impact (e.g. caused by an airbag) and is characterised by a transverse fracture through the ulna (see Figure 7.8). Humerus fractures result mainly from direct impact, but can also occur without any contact. Some cases are reported in which muscle forces as involved in overhand throwing caused the humerus to fracture [Levine 2002].

Surveying vehicle crashes in the UK, Frampton et al. (1997) analysed upper extremity injuries of car occupants whose vehicle were not equipped

with airbags. It was found that 86% of all upper extremity injuries were at AIS1 level (minor abrasions, contusions, lacerations). Hence, 14% formed AIS2+ injuries of which most injuries were fractures whatever the collision type. In frontal collision, forearm fractures were observed most frequently. Shoulder injuries were mainly found in struck-side crashes and rollovers. Clavicle fractures were identified to be the most frequent shoulder injuries. Humerus fractures, were found in struck-side crashes but were not common in frontal and rollover crashes. Hand injuries were recorded in some frontal collisions.

Investigating a sample of 540 crashes where the driver airbag deployed, Huelke et al. (1997) found a total of 34% of drivers sustained AIS1 upper extremity injuries and 3% sustained AIS2+ injuries to the upper limb.

Kuppa et al. (1997) found an increase from 1.1% to 4.4% in the occurrence of upper extremity injuries of severity AIS2+ as a result of airbag deployment. In contrast, Segui-Gomez and Baker (2002) who compared vehicles from model years 1993 - 1997 to vehicles from model year 1998 - 2001, noted a reduction of upper limb injuries in frontal crashes since the introduction of depowered airbags.

In summary, the following causes for upper extremity injuries were identified in the studies mentioned above:

- direct contact to airbag
- contact to interior of vehicle (including intrusions, e.g. in side impacts)
- contact of the arm with an interior part of the vehicle as a result of the arm being flung by the airbag
- inboard limb injuries due to contact with another occupant sitting next

Furthermore, it was observed that clavicula fractures may be caused by the seat belt diagonal section lying across the outboard shoulder and thus transmitting the belt loads transversely across the clavicula.

Additional studies indicate that women are at higher risk in sustaining a AIS2+ upper extremity injury [e.g. Bass et al. 1997, Schneider et al. 1998, Atkinson et al. 2002]. It is hypothesised that this is caused by the following factors: 1) women have generally smaller bones resulting in lower ultimate bone strength, 2) women experience an age-related loss of bone mineral density, 3) women are generally shorter in stature and therefore drive closer to the airbag system incorporated in the steering wheel, 4) young adult women tend to add bone to the endocortical surface, in contrast to men who add bone to the periostal side, resulting in a lower resistance to bending [Schoenau 2001].

Finally, it should be noted that the occurrence of airbag induced upper extremity injuries depends, of course, on the characteristics of the airbag. The term aggressiveness is used to describe the influence of airbag design

related parameters such as module (cover) design, pressure-time history, seam design and bag folding pattern. The aggressiveness is determined on a relative basis to assess the injury risk between different systems using devices like the Research Arm Injury Device (see section 8.4).

8.3 Impact tolerance

Early work by Weber (1859) and Messerer (1880) determined the load and moment required to produce failure in the bones of the human upper extremities. These studies remained the major reference data until in recent years upper extremity injuries received more attention again. Several research groups addressed the biomechanical response of upper limbs gaining additional data by performing further impact testing. Table 8.1 summarises tolerance values for the humerus reported in the literature.

Concerning forearm fractures, Bass et al. (1997) performed cadaver tests in which ulna nightstick fractures and multiple fractures were observed. The results suggest that the humerus position, the forearm pronation angle and the forearm position relative to the airbag module affect the risk of injury from airbag deployment. Furthermore, it was concluded that there is a forearm strength above which the risk of injury is low, even if the forearm is positioned in front of the airbag module. These findings support the

Table 8.1 Failure tolerances for the humerus.

humerus				reference
bending moment		shear force		
male [Nm]	female [Nm]	male [kN]	female [kN]	
115	73			Weber (1859)
151	85			Messerer (1880)
157	84	1.96 (overall)		Kirkish et al., (1996)
230	130	2.5	1.7	Kirkish et al., (1996), scaled to 50%ile male and 5%ile female
138				Kallieris et al. (1997)
	154			Duma et al. (1998a)
217	128			Duma et al., (1998b), scaled to 50%ile male and 5%ile female

hypotheses that women are at higher risk of sustaining upper extremity injuries.

Investigating the human forearm under a dynamic bending mode, Pintar et al. (1998) determined that the mean failure bending moment for all (male and female) specimens was 94 Nm. However, the bending tolerance of the forearm was found to be highly correlated to bone mineral density, bone area and forearm mass. Consequently, the study suggests that any occupant with lower bone mineral density and lower forearm mass is at higher risk to sustain a fracture.

Cadaver tests by Duma et al. (1998b) addressing the influence of the impact direction showed the forearm to be 21% stronger in supinated position (91 Nm) than in a pronated position (75 Nm). Conducting additional tests with female forearms in the pronated position and scaling those results to match the 5th percentile female geometry, a tolerance value of 58 Nm was obtained. Given that the forearm is typically pronated in the driving position, the value obtained from this weaker pronation position is meant to represent a conservative injury threshold.

The difference between static and dynamic impact was analysed by Begeman et al. (1999). Bending tests of the forearm were performed both quasi-static and dynamic by using a drop weight which resulted in a loading rate of approximately 3 m/s. Fracture of the ulna or the radius occurred with an average dynamic peak force of 1370 N and an average moment of 89 Nm. Static fracture loads and moments were approximately 20% lower. Nightstick or simple fractures were the most common type of failure. Differences between the radius and the ulna were not significant. In contrast to the work by Pintar et al. (1998) a correlation of the failure moment with age, cross-sectional properties, bone mineral content or moment of inertia was not found. As tests with one broken bone still showed a high failure moment, the authors suggest that other tissues may play a significant supportive role.

Regarding the elbow, Duma et al. (2001) observed that elbow injuries are caused not only by an axial force but also by a force that acts vertically relative to the horizontal forearm. Hence, a linear combination of the elbow axial and shear force showed a significant correlation to elbow injuries. Carrying out further cadaver tests, Duma et al. (2002) predicted, for the 5th percentile female, a 50% risk of elbow fracture at a compressive load of 1780 N at an elbow angle of 30° superior to the longitudinal axis of the forearm.

8.4 Injury criteria and evaluation of injury risk from airbags

To date neither governmental bodies nor consumer test organisations have released definite guidelines or regulations on how the risk of sustaining upper limb injuries in automotive accidents is to be assessed. There are no conclusive injury criteria or test protocols implemented yet.

However, Hardy et al. (1997, 2001) presented the concept of the Average Distal Forearm Speed (ADFS) to assess the risk of forearm fractures. Based on static and dynamics airbag deployment tests using cadavers as well as other test devices (i.e. Hybrid III dummy, RAID, SAE arm, for details see below), it was concluded that the distal speed of the forearm which is a function of both the forearm mass and the forearm proximity to the airbag module is a good predictor of the likelihood of forearm fracture. Scaling the results measured to the 5th percentile female geometry, it was found that a ADFS value of 10.5 m/s corresponds to the 50% probability of fracture. Furthermore, the ADFS decreased linearly with increased distance from the airbag module.

Further studies investigated the possibilities on how upper extremity injuries can be considered in the evaluation of crash tests and on how the aggressiveness of airbags with respect to upper limb injuries may be assessed.

To determine potential forearm fractures due to static deployment of driver airbags, Saul et al. (1996) found that the measurement of meaningful accelerations and bending moments was feasible using a specially instrumented 50th percentile male Hybrid III dummy arm. The arm correlated with airbag aggressiveness, proximity to the airbag module and relative position of the arm with respect to the airbag module.

Mounting the so-called SAE arm, which represents a specially designed and instrumented 5th percentile female upper extremity, to an Hybrid III dummy, Bass et al. (1996) determined that a forearm moment of 61 Nm ± 13 Nm represents a 50% risk of a ulna/radius fracture. The 50% risk of both a ulna and radius fracture corresponded to a dummy forearm moment of 91 Nm ± 14 Nm.

To enable reproducible evaluation of airbag aggressiveness, the Research Arm Injury Device (RAID) was developed [Kuppa et al. 1997]. Using the device to investigate the interaction between the deploying airbag and an upper limb close to it, it was found that the orientation of the forearm with respect to the airbag module and the distance between forearm and airbag module were significant parameters with respect to the measured peak

bending moment.

Despite the possibilities offered by the different arm test devices, inconsistencies between test objects were noted. Differences in the interaction of the shoulder-arm region with the side airbag were recognised by Kallieris et al. (1997) when performing static deployment test using cadavers and the Hybrid III dummy. Comparing the results of static side airbag deployment tests conducted with the Eurosid-1, a instrumented Hybrid III arm and cadavers, also Sokol Jafredo et al. (1998) observed great differences in the kinematics of the upper limbs. No correlation between the forces measured on the dummy and the cadaver could be established. Duma et al. (1998a) recorded kinematic differences between the SAE arm mounted to an Hybrid III female dummy and cadavers, but the moments recorded in the cadaver and the dummy were similar.

8.5 Upper extremity injuries in sports

As already described in previous sections, also in sports the upper extremities are prone to fracture, luxation, dislocation, (partial) rupture of tendons and ligaments as well as injury to muscles. Particularly the various joints of the upper extremities are at risk. The four joints of the shoulder girdle (Figure 8.3) together with elbow, wrist and hand allow for a wide range of motion and complex motion patterns to be performed. The ability of a joint to resist dislocation is directly related to its inherent stability. What a joint gains in mobility, it sacrifices in stability. This holds especially
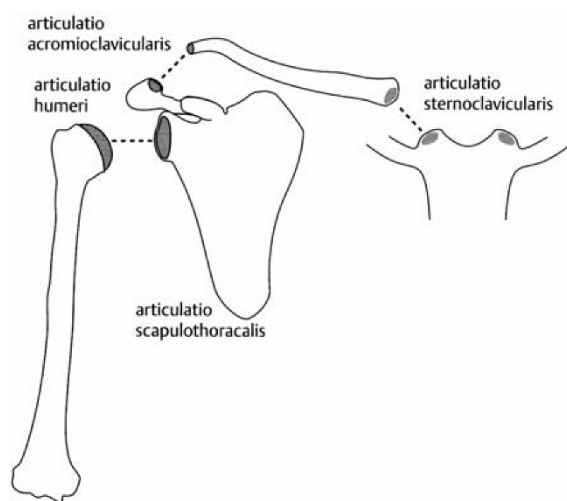
Fig. 8.3 Joints of the shoulder girdle [adapted from Brinckmann et al. 2002]

true for the shoulder which is prone to dislocation due to its relatively poor bony fit and limited supporting musculature.

Injuries involving the shoulder are a common consequence of sports participation and are estimated to comprise between 8% and 13% of all athletic injuries [Ong et al. 2002]. Sports with significant overhead demands such as tennis, baseball, volleyball and swimming often produce repetitive overuse syndromes. In contrast, injuries encountered in American football, hockey and other contact sports are often the result of direct trauma, e.g. fracture of the clavicle caused by a fall on the shoulder.

Traumatic shoulder injuries further include anterior glenohumeral instability (dislocation) due to a blow to the shoulder in the abducted and externally rotated position and, although less frequent, posterior glenhumeral instability which can for instance result a heavy frontal shoulder charge in field games. Acromioclavicular sprain may be initiated by direct or indirect forces that tend to displace the scapular acromion process from the distal end of the clavicle. Furthermore, injury to the rotator cuff muscles or the aromion is observed as a result from force transmitted along an adducted arm pushing the head of the humerus against the coracoacromial arch.

Particularly in sports with overhead activities shoulder overuse injuries are common, frequently involving to the tendons of the rotator cuff muscles (e.g. tendinitis in weight lifting). It is suspected that the shape of the acromion, i.e. whether it is flat, curved or hooked, influences the prevalence of such injuries. Similarly it is suspected that impingement syndrome is associated with acromion shape. Here impingement syndrome refers to arm abduction that results in suprahumeral structures (most notable the supraspinatus tendon and the subacrominal brusae) being forcibly pressed against the anterior surface of the acromion and the coracoacromial arch [Whiting and Zernicke, 1998]. Further causes for shoulder pain are related to biceps tendon disorders (e.g. rupture).

Due to the fact that the elbow is much more stable as the shoulder it is more than three time less likely to become dislocated [Whiting and Zernicke, 1998]. The bony structures of the elbow, however, are prone to fracture from direct blows as well as from indirect loading. Figure 8.4 summarizes different types of fracture depending on the mechanical impact.

Furthermore, particularly in racquet sports and sports involving throwing, the elbow often sustains overuse injuries. These include epicondylitis, tendonitis, myotendious strain and osteochondrosis of which epicondylitis is the most common. As a result of repeated loading existing microdamage increases, a progressive tissue degeneration is observed until
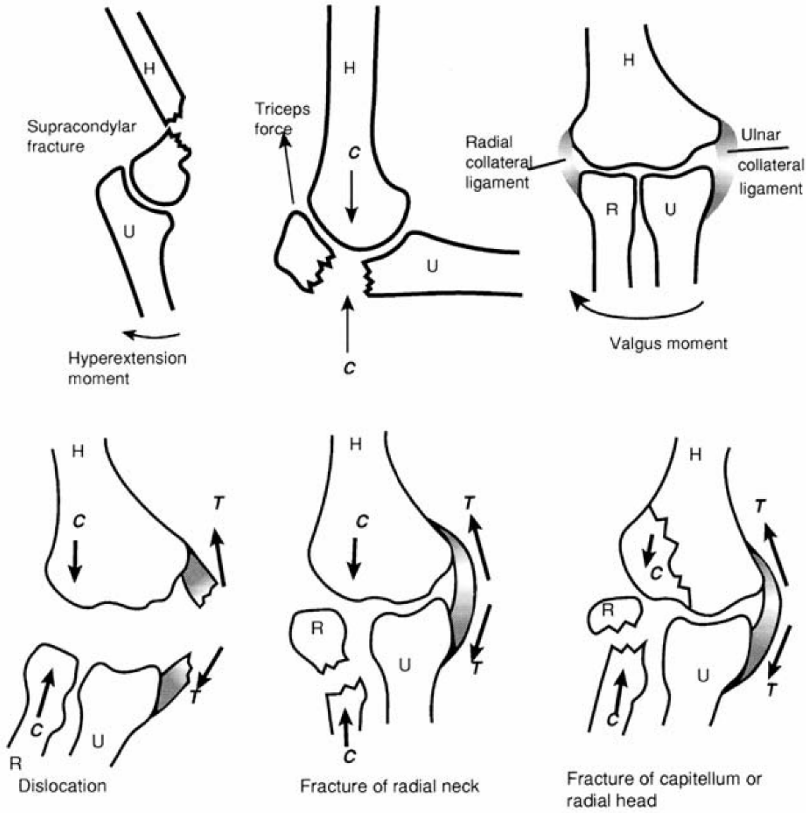
Fig. 8.4 Elbow fractures due to hyperextension moment, axial compression abduction moment (top from left to right), dislocations and fractures from a combination of abduction and hyperextension loading (bottom from left to right). H denotes for humerus, R for radius, U for ulna, C for compression and T for tension [adapted from Barlett 1999].

the tissue eventually exhibits inflammatory response. Lateral and medial epicondylitis are differentiated. Lateral epicondylitis is a degenerative condition of the tendon fibers that attach on the bony prominence (epicondyle) on the outside (lateral side) of the elbow. These tendons are responsible for anchoring the muscles that extend or lift the wrist and hand. A large number of tennis players are affected by lateral epicondylitis (hence it is also called tennis elbow). In racquet sports several causes are suspected to abet lateral epicondylitis including bad technique (particularly in backhand strokes), off-centre ball contact, grip tightness and racquet vibration [Whiting and Zernicke, 1998]. Medial epicondylitis in comparison affects the flexor tendon origin on the medical epicondyle and

is experienced by throwers, golfers and tennis players (forehand and service stroke).

Concerning forearm and wrist, fractures of the distal radius are most common in sports. Loading on the outstretched arm with (hyper-)extended hand, e.g. a fall in inline skating or snow boarding, induces compression forces that can lead to fractures. Several classification systems have been proposed to describe distal radial fracture using either clinical (radiological) criteria or characteristics based on injury mechanism similar to the classification described in section 7.2 [see e.g. Whiting and Zernicke, 1998]. In the context of investigating radius (and ulna) fractures, the ulnar variance is often referred to. Ulnar variance is defined as the ratio between the length of ulna and radius, i.e. it characterises the length difference between the articular surfaces at the distal radial-ulnar joint. If the two bones are of the same length, the ulnar variance is zero. Positive ulnar variance implies that the ulna is relatively longer than the radius, as determined from a radiograph at neutral rotation. Most often a person shows a small negative ulnar variance, with the radius taking approx. 80% of compressive forces submitted via the hand [Whiting and Zernicke, 1998]. A general correlation of the ulnar variance to the fracture risk is, however, not (yet) established.

Investigating the loading pattern to the wrist during pommel horse exercises Markolf et al. (1990) found that - depending on the exercise performed - peak forces up to twice the body weight and loading rates up to 219 times body weight per second can result. Furthermore, in young gymnasts, repetitive injury to the radial epiphysis prior to skeletal maturity is suspected to result in premature closure of the growth plate leading to an ulnar variance. Commonly reported wrist injuries and overuse syndromes in gymnasts are therefore not surprising, but general guidelines on how much loading the wrist may withstand are not available today due to the influence of too many individual factors.

To prevent wrist fractures various designs of wrist guards are available. These guards mainly aim at transferring loads from the hand to a larger area of the lower arm in case of a fall; additionally they prevent abrasions. The evaluation of the protective potential of wrist guards, e.g. used in in-line skating or snowboarding, has received significant attention, but is discussed very controversially. While some studies report a benefit from wearing wrist guards, others did not find an injury reducing effect [e.g. Schieber et al. 1996, Giacobetti et al. 1997, Greenwald et al. 1998, Staebler et al. 1999, Ronning et al. 2001, Hwang et al. 2006]. In experimental studies the incidence rate of fracture was investigated by loading cadaveric arms (with and without wrist guard) in conditions representing a fall on the

outstretched arm. Generally a wide spread of data was observed. Giacobetti et al. (1997), for instance, who tested 20 arms, report an average force to fracture of an unprotected arm of 2245 N (ranging from 1470 - 4116 N). The use of different experimental set-ups, the use of rather old cadaveric tissue as well as the use of different wrist guard designs makes it difficult to directly compare the results. However, from a biomechanical point of view, dynamic testing [e.g. Greenwald et al. 1998] is favorable over the quasi-static evaluations presented.

With respect to the hand, mostly fractures of the metacarpals and phalanges along with sprain and rupture of the collateral metacarpophalangeal and interphalangeal ligaments occur, particularly in contact sports [e.g. Bartlett 1999]. Although injuries to the hand diminish an athlete's ability to perform and often result in a loss of playing time, many of these injuries may be treated effectively nonoperatively such that the athlete can return to sport rapidly [Snead and Rettig, 2001].

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9 Impairment and injuries resulting from chronic mechanical exposure

An accident is defined as a violent, unsuaul and possibly harmful event which occurs suddenly and unexpectedly and is mostly of a short duration. Persons involved in an accident can in general not or only insufficiently react in order to prevent injury. The term "chronic", in contrast, implies that a process extends over durations which are long in comparison with typical accidental time intervals. Accordingly, physical and mental reactions of the persons involved always occur and cannot be neglected. Of primary importance is however the fact that under conditions of chronic mechanical overexposure, impairment of health and injury may result from an amount of mechanical load to which the body is exposed or of functional misuse in the course of some physical activity which is well below an acute tolerance limit for an individual (such as is described in the other chapters of this book), but whose effect is aggravated and outperformed by an extended duration during which it acts. The exposure may thereby often be interrupted and limited to regularly or irregularly occurring time intervals, e.g., to training periods (sports) or work assignments (jackhammer) which may however extend over years. A single, isolated exposure as such is usually mostly harmless or allows for a straightforward recovery. As a result, causation, mechanisms, patterns, tolerances, prevention and mitigation of adverse health effects are quite different from what is observed in accidents.

A distinction between injury and disease is often not well defined. Long-term sequelae of exposure to potentially harmful mechanical loads associated with a certain profession, e.g., in construction work, are considered as occupational illnesses and treated as such. Or, the Repetitive Strain Injury (RSI) syndrome which is observed among other in tennis, is denoted a disease rather than an injury despite its name because it manifests itself as a form of impairment representing the final clinical result of a long sequence of numerous microscopic injuries.

Mechanical exposures which may cause health problems appear in quite various forms. The spectrum of mechanical loading scenarios thereby reaches from stochastic sequences of single, isolated impacts during certain periods of work, training or games, such as occurs for instance in boxing, to continuous vibrations, e.g., loud sound. Of comparable variety are the characteristics, professions, activities and social environments of the involved persons. Accordingly, reporting and documentation practices, intervention of authorities, legal subsumption, regulations, liability issues and insurance coverage are quite diverse and different from procedures which come into play after an accident with injury. Likewise, government agencies in charge of chronic disease control - regardless of the origin of the impairment - have in general no relation with traffic safety or general accident and risk management.

Chronic mechanical overexposure, overuse or functional misuse can manifest itself in a variety of different forms and severity:

- A worker using a drill hammer routinely can suffer from TFCC (triangular fibrocartilage complex) lesions.
- After a strenuous hiking tour, our feet may be covered by painful, but harmless blisters.
- A promising career in sport may be terminated by too much and inappropriate training.
- Repeated exposure to loud music in a recreational facility can induce a permanent hearing loss.
- Lower back pain leading to partial disability can result from routine household work in an unphysiological position.
- Hypertension forces the heart to perform an increased, ultimately harmful work load to maintain a physiological cardiac output.
- The heart of a bicycle or rowing champion which develops hypertrophy resulting from intensive training needs careful attention and a well planned period of regression after discontinuation of the competitive activity.

In traffic, safety is almost exclusively devoted to acute situations, while potentially harmful chronic loading is generally controlled by comfort-oriented and ergonomic design of the user environment. Occupational illnesses and injuries, in particular those which are due to chronic mechanical exposure, have in turn extensively been examined in the past because of their high socio-economic significance, political implications, insurance claims and workman's compensation programs. In industrialised countries, workplace safety is therefore governed by numerous regulations and subjected to rigorous control by government bodies and insurance companies. The situation with respect to sports-related impairment which is

Table 9.1 Number of accidents/injuries and illnesses along with cost due to sports, estimated for Switzerland in 2001 by the Swiss Federal Health Office. While direct cost are related to medical treatment, indirect cost are caused by absence from work, insurance administration, forensic expenditures, etc. The figures have to be considered in relation to the population of 7.5 million and a GNP of around 400 10^9 US$.

	Number of accidents/ injuries	Number of diseases	Direct cost [CHF]	Indirect cost [CHF]
Accidents/injuries and cost caused by sports	300'000		1.1 10^9	2.3 10^9
Benefit from sports-related physical activity		2.3 10^6	2.7 10^9	1.4 10^9
Illnesses due to lack of physical activity		1.4 10^6	1.6 10^9	0.8 10^9

likewise associated with enormous health cost (Table 9.1), is however quite different. Public awareness is mostly related to the health condition of star players in prominent teams rather than to health impairment resulting from sports activities of the general population. In professional sports, safety issues are primarily reflected in game rules, referee work and trainer education, while in general sports, countermeasures are often of an ad hoc nature, little systematic, limited to general recommendations or promoted by manufacturers of sports accessories. The reason might derive from the fact that most sports activities are of a recreational and voluntary nature. Again quite different, finally, are circumstances regarding health problems due to continued strenuous household work: This area is virtually unexplored from a scientific point of view.

Medical disciplines mostly involved in chronic diseases are rheumatology, orthopaedics, neurology, sports medicine, radiology, pain management. Due to the significant involvement of psychic and social factors, psychiatry is likewise of importance. While diseases considered here such as bursitis (inflammation of a bursa, a friction-reducing layer between skin, ligament or tendons and bony structures) are mostly due to external mechanical overloading over extended periods of time, mechanical overload can also be caused by the body itself, in that obesity, hypertension

or muscular dysbalance represent long-term risks in view of health impairment and premature death. Functional misuse, e.g. continued work in a bent position, is a further source for health impairment. The application of biomechanics in the quantitative sense treated in this book is difficult under circumstances exhibiting such a variety. "Simple" statistics, i.e., collection of numbers and monitoring of data over years prevail. This is not surprising since realistic experiments on long-term mechanical exposure can hardly be performed, analytical procedures suffer from missing basic knowledge and physiological and psychical reactions under conditions of continued pain exhibit enormous individual variations which cannot be disregarded. Accordingly, workplace safety issues are to a large extent limited to the prevention of acute injury.

Rehabilitation is an important issue likewise in cases of acute and chronic illnesses. Biomechanics are of important in this context insofar as physiotherapy, physical therapy, orthopaedic aids, wheelchairs, training devices etc. rely on the application of methods and on the design of apparatus where biomechanics have to be taken into account mostly on a quite basic, albeit important level. Since this is not directly related with trauma-biomechanics, however, these subjects are not further considered here.

9.1 Occupational health

Professions and working conditions which are associated with heavy and strenuous mechanical exposure are numerous and cover a wide variety of forms ranging from as different activities as underground mining to ballet dancing. The problem of long-term impairment appears in all of these professions, but it has to be treated differently in each case. Moreover, working conditions and standards to be maintained differ considerably from country to country. Activities of labour unions and NGOs as well as legal procedures with respect to early retirement, inability to work and workman's compensation are furthermore quite influential and subjected to different local political conditions.

A large number of regulations and status of health reporting plans in connection with strenuous and hazardous work is maintained by the Occupational Safety and Health Administration of the US Department of Labor (http://www.osha.gov). In other countries, similar organizations exist. In Switzerland, e.g., it is the partly government-controlled Accident Insurance Company (suva) with mandatory coverage of all "blue collar"

employees (i.e., including also low-risk professions) whose tasks includes workplace as well as leisure accident issues. From the many professions, a few exemplary remarks are made in the following..

- Construction work, mining, lumber and woodwork: These are the professions that come to mind first when a hard and strenuous work environment is addressed. Early retirement due to health problems and health-related absentism are in fact well known [Brenner and Ahern 2000] and have been recognized as such for a long time. While the risk of acute injury is approached by numerous regulations (helmets, ear protection, gloves, etc.) long-range impairment is difficult to control. For practical reasons, e.g., weights to be lifted cannot be limited because these are given by the materials to be used. Accordingly, lower back pain is widespread [Latza et al. 2002]. Cost associated with health problems along with the need for a continued increase in efficiency causes therefore a growing application of machinery.
- Nursing: Nursing is known as a profession which is partly associated with strenuous work and impairment. The situation was analysed comprehensively in the EU sponsored NEXT study (Nurses Early Exit Study). The physical environment, i.e., lifting, bending or work with noncooperative and aggressive patients was thereby recognized as a major problem.
- Professional dancing (classical ballet, break dance): Amenorrhea is a well known complication in females who are exposed to continued strenuous or irregular work environments such as professional ballet dancing or long-haul airline work. Associated loss of bone mineral content cannot systematically be compensated by physical training [Warren et al. 2002]. Not surprisingly, the ankle joint is mostly prone to chronic impairment in ballet dancers [Rand 1999]. A comprehensive overview over risks associated with professional dancing is given in [Dance Medicine - The Dancer's Workplace, Unfallkasse Berlin].
- Noise: Hearing loss is a common outcome of extended exposure to loud noise. In some cases, enormous sound levels are reached even using tiny music players with ear phones (Table 9.2). Since the advent of high power music equipment with widespread use the problem aggravated substantially because a large number of young people is affected. It should however not be forgotten that problems exist also under traditional conditions: The maximal sound level to which a musician in the Wagner orchestra is exposed approaches 140 dBA. The risk of hearing loss is strongly correlated with the exposure time. According to the Swiss Accident Insurance, 4 hrs./week of disco-music at 93 dBA (closed room) or 2 hrs./week of outdoor concert at 100 dBA are

considered tolerable; for lack of precise knowledge, a linear scale is assumed for extrapolation.

Table 9.2 Observed maximal sound levels in various circumstances (Source: The Safe Side, Wisconsin, USA, vol. VII, 2004).

	noise [dBA]		noise [dBA]
Firecrackers	125-155	Lawn mower	90-110
Concerts	120	Car horn	110
Gun shots	150-167	Jack hammer	113
Movie theaters	118	Hair dryer	90
Sporting events	127	Chain saw	110
Health club, aerobics studies	150	Personal stereos	105 - 120
Motor boats	85-115	Children's toys	135-150
Motor cycles	95-120		

9.2 Sports

9.2.1 Non contact sports

Overuse injuries account for approximately 50% of all injuries in sports [Wilder and Sethi 2004]. They are mostly due to physical overuse (overuse syndrome). Thereby, repeated microtrauma beyond the reparative abilities of the musculoskeletal system will eventually lead to macroscopic injury and clinical symptoms. "Repetitive Strain Injury (RSI) syndrome" is a further expression which describes this phenomenon.

Most often clinically observed health problems include:

- Tendinitis (painful inflammation of a tendon) is among the most common problems diagnosed in this context. It is furthermore aggravated by age-related degeneration that affects many of the large tendons in both the upper and lower extremity [Karamanidis and Arampatzis 2006]. This leads to an increased predisposition with respect to painful lesions during athletic activity. Changes of collagen

composition have been identified as a major reason [Riley et al. 1994]. Specific examples are rotator cuff tendinitis, tennis elbow (epicondylitis), and Achilles tendinitis, observed in activities such as running, overhead throwing and serving balls in tennis. It is important to note that these lesions are ultimately associated with unrecoverable degenerative processes rather than "only" inflammatory changes. The term tendinopathy may in such cases be more appropriate than tendinitis, which implies only inflammatory changes.

- Stress fractures are characterized by tiny cracks in bone often caused by repetitive overloading (such as in the feet of a basketball player who is continuously jumping on the court) [Snyder et al. 2006, Wilder and Sethi. 2004, Egol et al. 1998, Fredericson et al. 1997]. A reaction to such injury is demonstrated in Figure 1.2 where microcallus formation is shown. While such microscopic injuries - if sufficiently scarce - favour bone remodeling (our skeleton is totally replaced within 4 - 6 years under healthy conditions) [e.g. Martin 2003], continued extensive microcrack formation is deleterious. In case of adolescent athletes, furthermore, epiphysitis or apophysitis, i.e., growth plate overload injuries such as the Osgood-Schlatter disease (swelling and pain below the knee) or a process denoted as "shin splint" (similar symptoms) are observed [Wilder and Sethi 2004].

Due to the various, for sports typical extensive jumping and running activities, the lower extremities, in particular the feet, ankle [Valderrabano et al. 2006], sport shoes and the strucutre of the ground (stiff, elastic or energy absorbing) are of importance [e.g. Bartlett 1999]. Since typical and systematic loading scenarios can be defined in this context, a quantitative analysis in the design of sport shoes is possible [Reinschmidt and Nigg 2000, Stefanyshyn and Nigg 2000]. In case of running shoes, major aspects include pronation control and cushioning. For court shoes, in turn, lateral stability, torsional flexibility, cushioning and traction control have been identified as significant design parameters in view of injury prevention. A further problem associated (not only but especially) with sport shoes is onychodystrophy, i.e. mycosis (fungal infection) due to chronic mechanical irritation [Romano et al 2005]. The knee, in particular with jumpers, is likewise of concern [Tibesku and Pässler 2005].

Extensive recommendations can be found in order to prevent sports-related overuse injury [Niams 2007, NCSS 2007, bfu 2007]. Additionally, trainer education, mostly in professional sports, represents an important task.

9.2.2 Contact sports

Some forms of contact sports (boxing, kick-boxing, wrestling, martial arts) are potentionally prone to cause injury. It is the primary task of the referee(s) to enforce regulations and prevent acute serious injury (a ruptured eyebrow or lip does not represent a serious injury in this environment). Nevertheless, while a knock-out associated with unconsciousness as a single event may be harmless and recovery is, in general, rapid, continued exposure to violent blows is not. This is reflected, e.g., in the expression "slap-happy" which refers an (unsuccessful) boxer who suffers from partial dementia due to repeated blows to the head.

Boxing, kick-boxing, wrestling and martial arts are partly covered by strict regulations such as the ones issued by the World Boxing Association [WBA 2007] or the Fédération Internationale de la Lutte Amateur [FILA 2007]. Nevertheless, long-term sequelae of such sports are not primarily taken into account in these initiatives.

9.3 Household work

As mentioned above, long-term effects of household work are mostly unexplored in spite of the fact that unphysiological postures during kitchen work (low tables causing back pain) or cleaning activities (bent position over extended periods of time) are not uncommon. Impairments due to such circumstances are mostly treated by family doctors individually without significant amounts of insurance compensation. Nevertheless, some consumer organizations try to analyse the problem and maintain a consulting and recommendation service for the public.

9.4 Conclusions

Chronic diseases and injuries which are due to mechanical overloading, overuse or functional misuse represent a major social and socio-economic problem. In the industrialised countries, this problem is further aggravated by the increasing life expectancy associated with a growing percentage of aged persons. Yet, biomechanics as they are treated in this book has a limited significance in this area. Quantitative tolerance limits with respect to loading which would have to be defined as function of frequency and duration of loading hardly exist. This is primarily due to the fact that a quantitative procedure is mostly not possible in that the basic principles of

biomechanics, i.e., measurement, analysis and modelling of long-term mechanical exposure have hitherto not been applied widely and systematically. Accordingly, only qualitative recommendations, general rules, monitoring and statistical approaches prevail. In view of the high significance of chronic impairment, however, a great challenge exists for future biomechanics' research in this particular field.

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